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DISSERTATION

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„Biomechanical assessment in over-ground running“

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Abstract

The purpose of this study was to investigate the strength and limitations of a biomechanically grounded assessment in over-ground running. Traditionally, physiologic measures and their relation to performance are used, in order to assign training regimes for athletic runners. Running form is inspected visually or bound to occasionally events in the laboratory. However, there are propositions towards computer-aided training systems, which support athletes during their training. Current technology offers to make use of wearables—small devices gathering and distributing data. The thesis aimed at proposing a training system, which allows for observing runners live and on the Internet by remote located experts. Moreover, this 'Mobile Coaching' system should be able to send feedback to runners. Therefore, the literature and own research were exploited, in order to reveal criteria for the assessment and appropriate parameters. It appeared that stride parameters are related to performance and the injury risk. The effect of fatigue on both was investigated through the behavior of stride parameters and derived indexes. Based on this, kinematic markers for the evaluation of performance and the injury risk were established. While former studies were limited to the analysis of few strides, there was a need for a greater amount of strides, because the consideration of variability suggested to draw conclusions on neuromuscular properties. Striding further than the physiological and biomechanical optimum (over-striding) as well as a too low stride rate were identified as the main issues with recreational runners. Furthermore, a pedestrian deadreckoning approach was extended, in order to capture time series of stride parameters in a full-exhaustive middle-distance time-trail. As previous studies examined temporal parameters, this study also analyzed the stride length. The stability of the movements did not change as it was examined with the detrended fluctuation analysis. It was concluded that

the analysis of the time course and its variability allow for profiling runners and the provision of feedback. However, the analysis of stability might gain incentive value when several running trails are considered, in order to estimate long-term effects of fatigue on the neuromuscular properties.

Zusammenfassung

Die Dissertation hatte als Ziel die Grenzen und Möglichkeiten eines biomechanisch orientierten Trainingssystems für das sportliche Laufen zu ermitteln. Moderne Trainingsplanung basiert in erster Linie auf einer physiologischen Leistungsdiagnostik, um die sportliche Leistung zu verbessern. Obwohl biomechanische Parameter zur Beurteilung der Lauftechnik vorliegen, wird die qualitative Analyse eingesetzt, da sie einerseits einfach und kostengünstig durchzuführen ist und andererseits kaum Messsysteme für den Dauereinsatz vorhanden sind, so dass biomechanische Analysen in der Praxis an Untersuchen im Labor gebunden sind. Moderne Technologie bietet den Einsatz von „Wearables“ an, welche als miniaturisierte Systeme oder Sensoren es ermöglichen, sich nahezu rückkopplungsfrei in das sportliche Training zu integrieren, um ausgewählte Daten für das Generieren von Trainingsanweisungen zu gewinnen. Der praktische Teil der Dissertation war darauf ausgerichtet, ein solches Trainingssystem ("Mobile Coaching System") zu entwickeln bzw. vorzuzeichnen, welches ExpertInnen die Beobachtung von LäuferInnen in Echtzeit und von beliebigen Orten über das Internet und das Versenden von Rückmeldungen an den/die SportlerIn ermöglicht. Mit Hilfe einer Literaturanalyse und eigenen Untersuchungen konnte ein Zusammenhang zwischen den ermüdungsbedingten Veränderungen in den Doppelschrittparametern (Länge, Frequenz, Bodenkontaktzeit) und der sportlichen Leistung als auch dem Verletzungsrisiko hergestellt werden. Von diesen Veränderungen wurden kinematische Marker zur Anzeige des neuromuskulären Potentials (z.B. Effizienz des Dehnungs-Verkürzungszykls) des/der Laufenden abgeleitet. Im Gegensatz zu früheren Studien wurde eine beträchtlich höhere Anzahl an (Doppel-)Schritten ausgewertet, was die Aufarbeitung der (biologischen) Variabilität in den Parametern begünstigte und damit weitere Schlussfolgerungen auf neuromuskuläre

Eigenschaften zuließ. Schrittlängen größer als das physiologische und biomechanische Optimum bei zu geringen Schrittfrequenzen wurden bei FreizeitäuferInnen als mögliche Ansatzpunkte für eine Intervention festgestellt. Um Zeitreihenanalysen zu ermöglichen, wurde u.a. ein Ansatz aus der Passentenverfolgung (mit Inertialsensoren und GPS) erweitert. Dadurch konnten im Gegensatz zu vorangegangen Studien nicht nur temporale Parameter sondern auch die Doppelschrittlänge ausgewertet werden. Diese Zeitreihen wurden mit der monofraktalen Fluktuantionsanalyse (Detrended Fluctuation Analysis, DFA) untersucht, da diese eine Einschätzung der Stabilität der Bewegung erlaubt. Dieser Analyse zufolge änderte sich die Stabilität während eines 5-km-Zeitlaufs nicht. Es wurde geschlussfolgert, dass die Analyse der Zeitreihen, d.h. die Wechselwirkung zwischen den Parametern und der Variabilität, eine Einschätzung des Zustandes der LäuferInnen erlaubt, um weitere Trainingsanweisungen zu generieren. Ob sich die Fluktuationsanalyse jedoch für die Analyse von Langzeiteffekten durch Ermüdung auf neuromuskuläre Eigenschaften anbietet, wird weiterführenden Studien zur Exploration empfohlen.

Gewidmet meinem Opa

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Contents

List of Figures

List of Tables

Glossary

BH Body height

BSN Body sensor network

BW Body weight

CT Ground contact time defined as from initial contact to toe-off or as duration of the stance phase.

CV Coefficient of variability—mean divided by STD

DFA Detrended fluctuation analysis

FP Footpod, uni-axial accelerometer attached to the hindfoot

FT Flight time of the leg during a stride

GPS Global Positioning System

HR Heart Rate

IMU Inertial measurement unit

IMU/GPS Combines IMU and GPS

IMU/RADAR Combines an IMU and a radar device, in order to determine stride parameters based the traveled distance over time and SR.

LL Leg length

MC system Mobile Coaching system

PDR Pedestrian Deadreckoning

PvC Pervasive Computing

SL Stride length

SR Stride rate, cadence, or frequency

SS Stride speed

ST Stride time

STD Standard deviation

UC Ubiquitous Computing, another expression for PvC

VT_2 Anaerobic ventilatory threshold

ZUPT Zero-Velocity-Update. The velocity of the foot is assumed to be zero at one time during stance.

1

Introduction

1.1 Background and Motivation

The PhD thesis on hand is in the research area of Mobile Coaching (MC) at the department of biomechanics, kinesiology and applied computer science at the University of Vienna. Mobile Coaching develops methods and devices to provide athletes, coaches and other people involved in athletic activities with information about their performance and suggestions for improvement. These persons interact with an MC system, which provides the athletes with feedback in as close to real-time as possible. MC is also a part of training science, but furthermore uses technical devices extensively. The era of Pervasive Computing (PvC) paved the way for MC since technological developments have recently offered a variety of sensors which can be mounted on the athlete, or training equipment with hardly any disturbance. In this light, MC has become a multidisciplinary and integrative approach to sports sciences.

The increasing interest in building MC systems can be drawn from the manifold examples of elite, amateur, leisure, recreational, and rehabilitative training. The research area of the supervising department mainly covers rowing, table-tennis, bicycling, weight-bearing sports—and running (Baca *et al.*, 2010; Novatchkov & Baca, 2012; Bichler *et al.*, 2012; Baca & Kornfeind, 2008; Kornfeind & Baca, 2008), whereby the goal of this thesis was to demonstrate how a biomechanical assessment can support runners in an MC environment.

As depicted in fig. 1.1, the challenge of this thesis was to overcome the complexity arising from the diversity of the research areas PvC, MC and the biomechanics of running assessment. The literature review gave insights of the variety of approaches undertaken, from which it was decided to design a biomechanical-based MC systems for runners. Nevertheless, the principals of training science according e.g., to Carl (1989) recommended to follow a procedure that figures out goals for the athlete, then compares these to the current state of the athlete and finally applies appropriate methods of training by repeating until the goals are achieved by an appropriate selection of exercises and sets of duration, breaks, intensity and corrections, see fig. 1.2. This line of action implies that there is a conception of what is desirable and how to evaluate the training process. Furthermore, relevant state and control parameters have to be chosen and determined. In practice, biomechanical based analyses rely on measurement equipment,

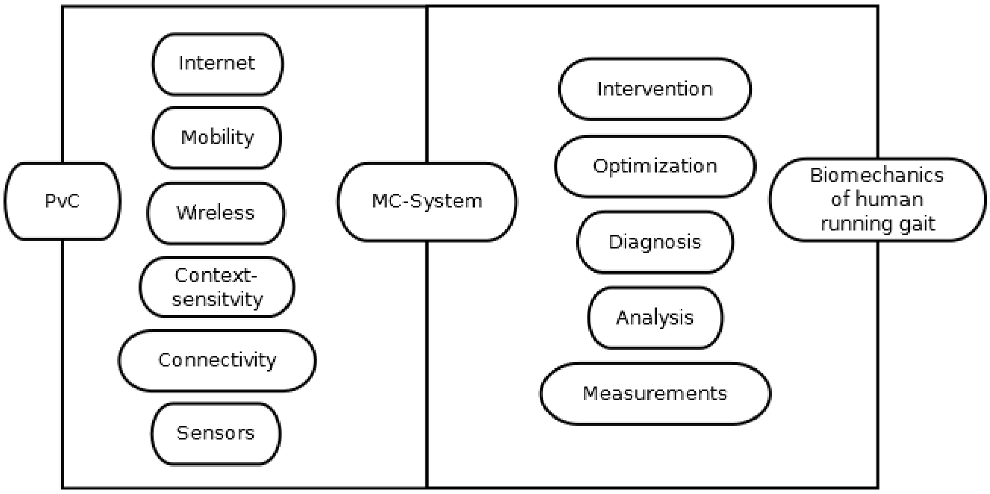


Figure 1.1: Relation between the research areas: PvC, MC and biomechanics of human running gait

resulting in quantitative measures. The feasibility of measuring desired parameters is limited by the technology available. Pervasive (PvC) or Ubiquitious Computing (UC) has opened new doors for novel tools towards the analysis and diagnosis of running gait in an outdoor scenario by an MC system.

PvC is the trend of the technical development when computing devices get smart, tiny and even invisible—the calm technology—as Weiser (1991)[1] called it. It is not just the increase in power of its components or the technical alternatives with less energy consumption, rather PvC is characterized by an interconnection of those technical devices and an interaction of humans with them. A wide range of applications have been developed from 'ambient assisted living' to telemedicine (Chaaraoui *et al.*, 2012; Fong *et al.*, 2011; Wartena *et al.*, 2009; Djumanov *et al.*, 2008; Otto *et al.*, 2006). Thereby, sensors and sensor networks play a crucial role in a context sensitive environment. Within this thesis, the focus of selecting proper sensors was mainly based on properties like size, sensitivity, accuracy and transmission of signals, so that they met the requirements of running. The handling of the higher velocities of running in contrast to walking revealed itself as the most outstanding demand challenging sensors and quantitative determination.

[1]Weiser is considered to have coined the expression *Ubiquitous Computing*.

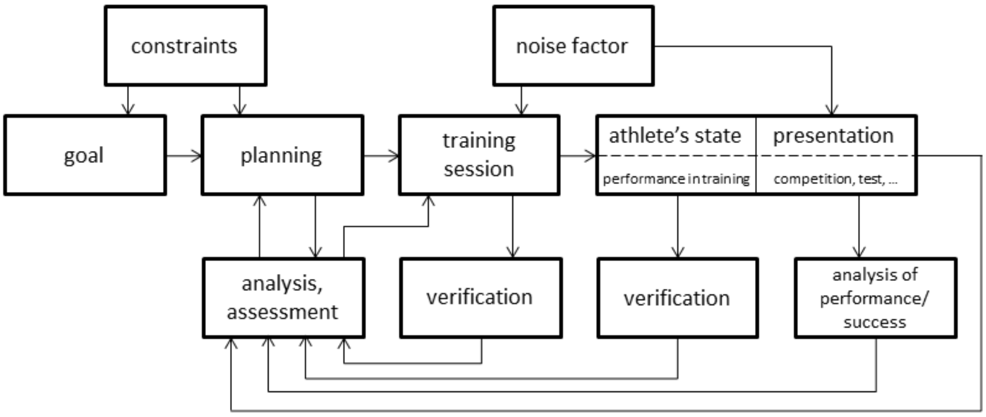


Figure 1.2: Evaluation of a training process according to Carl (1989)

The MC system's underlying approach used in the thesis was to evaluate training parameters automatically and send feedback to the athlete or athletes based on intelligent analysis of the training data. Remotely located experts should have the ability to inspect and influence the training, too. Intelligence in this context means that the MC system is aware of the athlete's needs and adapts its training advice accordingly (Baca *et al.*, 2009) by taking into account a variety of information about the athlete or athletes, and the training, see fig. 1.3.

In the past decade, a commercial trend has been kicked off which enables logging training session (Zong, 2008) via mobile Internet devices like smart-phones, PDAs, ultra mobile PCs or tablet-PCs. Most of the reviewed commercial smart-phone based services for running survey the position of the runner and the heart rate (HR). Some of these services enable others monitoring the athlete in real-time e.g., via a web interface or sharing training data in social networks.

In research literature, papers could be uncovered having versatile proposition towards an MC system. For example, Kugler *et al.* (2012) introduced a biofeedback system collecting kinematic and kinetic data as well as biosignals from electrocardiogram (ECG) and electromyogram (EMG). A body sensor network (BSN) managed the sensor nodes and the data transmission between the sensors and the smart-phone. Lopez-Matencio *et al.* (2010) named their version of an MC system an 'ambient intelligence assistant for running sports'. They emphasized the integration of environmental data like temperature and wind speed into the estimation of the performance of the runner. The heart rate and the oxygen saturation were employed to describe the runner's physiologic state. According to the state and training goal of the runner, suitable tracks were proposed during the run by feedback messages. In 2012 Tampier *et al.* presented a real-time feedback system for running. This MC system provided the runner with an optimal speed according to a minimal duration of the run, the profile of the track, fitness level of the runner, and the fitness or fatigue state (Perl & Endler, 2011) by implementing the physiologic inspired load-performance-meta-model "*PerPot*" (Perl, 2001).

Heart rate variability (HRV) shows promise in the field of physiology (Canovas, 2011; Hansen *et al.*, 2004; Saalasti, 2003). Goldberger (2002) emphasized the "extraordinary complexity" of the "physiologic systems" at all and referred to the "growing interest" in analyzing physiological time series by "applying concepts and techniques from physical statistics, including chaos theory, to biomedical problems ranging from the molecular level to the level of the entire organism". For example, the analysis of HRV assists the estimation of the stress level of the runner—before and after the training—and detecting phenomena like overreaching and overtraining (Mourot *et al.*, 2004; Aubert *et al.*, 2003; Pichot *et al.*, 2000; Lehmann *et al.*, 1991). Psychological issues were included, for example, by taking note of subjective rated perceived exertion (Eskofier *et al.*, 2008). Music in the field of feedback systems for running enjoys popularity too (Oliver & Kreger-Stickles, 2006; Oliver & Flores-Mangas, 2006; Elliott & Tomlinson, 2006). More examples for MC coaching systems close to running can be found e.g., see Preuschl *et al.* (2010), Harms *et al.* (2010), Buttussi & Chittaro (2008) or Miller (2001).

However, the versatile implementations of MC systems for running available used GPS-data derived parameters, physiological and psychological input or music approaches. Despite the presence of MC systems being prepared for kinetic and kinematic data (Kugler *et al.*, 2012; Harms *et al.*, 2010), a pure biomechanical approach was not found. For this reason, the thesis made an effort to investigate the strengths and limitations of a biomechanical analysis and put forth the question:

Which biomechnical data needs to be observed and how can this data be

Figure 1.3: The desired structure of an MC system for runners. The smart-phone of the runner should gather data from the sensors (red points) via a BSN. The preprocessed data will be sent to a server via the Internet and stored in a database. The analysis of the actual training should be carried out by experts e.g., sport scientists or biomechanists, or server-side routines. Finally, the runners should be supported by feedback.

used to give feedback to support the training?

"Improvements in running economy have traditionally been achieved through endurance training" (Bonacci *et al.*, 2009, p.903). Indeed, further review in the field of performance diagnostic in running gave the impression that runners mainly take advantage of physiologic based aerobic or anaerobic tests followed by strength, power, flexibility, and agility tests, in order to evaluate their performance and draw conclusions for their training (Oliver & Stembridge, 2011; Milne, 2006; Bangsbo & Sjogaard, 2001). In contrast, kinematic or kinetic assessments do not serve this purpose. Though biomechanical measures for assessment in daily training were not found, qualitative running analysis has been approved in track and field sports for a number of reasons (Masci *et al.*, 2012; Bartlett, 2007; Knudson & Morrison, 2002). The assessment of the running technique and the state of the runner during the training relies on the coach's and athlete's subjective perception and the static control parameters gathered by the performance diagnostic. Furthermore, running movement is highly individual, thus running styles differ strongly. Nonetheless, Dillman (1975) and Cavanagh & Williams (1982) suggested that energetically optimal stride length compared to oxygen uptake can be expressed in terms of leg length or body height.

The human gait of walking and running has been attracting scientists since the first evidence of human culture. Aristoteles is perhaps one of the most famous ones (Novacheck, 1998). Since the fifth century, "researchers attempted to model the human musculoskeletal system" (Godfrey *et al.*, 2008). Although the biomechanics of running has been well studied to date (Dugan & Bhat, 2005; Anderson, 1996; Cavanagh, 1990; Mann & Hagy, 1980; Saunders *et al.*, 2004; Hunter *et al.*, 2004; Novacheck, 1998; Adelaar, 1986; Dillman, 1975), there are few practical guidelines based on quantitative analysis for daily use—aiming at the diagnosis of the running technique and an optimizing intervention. However, sophisticated methods of motion analysis are bound to laboratory environments or are too expensive.

Distance runners and coaches may follow certain (kinematic) models e.g., such as Marquardt (2011), Traenckner (2007), Romanov & Fletcher (2007), and Neumann & Hottenrott (2005). Differences in those guidelines can be found; for example, the "Pose method" of running introduced by Romanov & Fletcher (2007) has a very different concept than the aforementioned authors. The major differences in posture, initial foot contact, stride length and rate are involved in the etiological theme of movement science. In this view, the process of learning the gait of walking and running appears in the light of evolution and the circumstances of the human environment and equipment. For example, shoes, inserts, cushions and unnatural terrains such as tartan tracks or asphalt roads have sufficient influence of the etiology of running. In general, recommendations consider the non-fatigued runner and direct their attention to an optimal kinetic chain. The criterion of optimization can be e.g., the oxygen uptake, energy expenditure, forces at the patella, effectiveness and efficiency of the stretch-shortening cycle, or 'optimal' running kinematics.

There is a growing body in literature studying the kinematics of running. Stride parameters appeared to be determinable within a framework of a MC system. Furthermore, the stride parameters length (SL), rate (SR), its inverse the time ST, (ground) contact-time (CT), and flight-time (FT) have been found of importance to characterize athletes and select appropriate exercises (Incalza, 2007) to improve performance (Landers *et al.*, 2011) or/and reduce the injury risk (Fletcher *et al.*, 2010), and furthermore being beneficial for feedback provisions (Incalza, 2007). For example, as early as in 1976

Nelson & Gregor, attempts to evaluate a training program for distance runners with respect to the adjustments in these stride parameters revealed individual but also group effects. Adjustments during the run—the deviation from the (individual) norm—may play an important role for the investigation of the runner's state, see e.g., Chan-Roper *et al.* (2012), Morin *et al.* (2011a), Schornstein (2011), Morin *et al.* (2011b), Hanley *et al.* (2011), Hanley & Smith (2009), Incalza (2007), Hanley & Mohan (2006), Dallam *et al.* (2005), Hardin *et al.* (2004), Derrick *et al.* (2002), Dutto & Smith (2002), Hay (2002), Kyroelaeinen *et al.* (2000) and Mizrahi *et al.* (2000). All the aforementioned studies stated a change in some biomechanical parameters (under full exhaustion), but are not consistent. For example, the stride length did not decrease over time in all studies. Such differences might be also due to different race distances. Nevertheless, it can be reported that significant adjustments of biomechanical parameters have been detected. For example, most of the reviewed studies in over-ground running mentioned the effect of fatigue on the contact time, which increased slightly. This is likely due to the attenuated efficiency of the stretch-shortening cycle. Kinematic adaptations occur under fatigue and can be linked not only to a change in metabolic conditions but also to a change in the capacity of motor control e.g, the number of recruited motor units.

Hence, the recognition of fatigue and its associated potential neuromuscular stress might be suitable to draw conclusions, in order to estimate the state of the runner within an MC system, and thus may help to select appropriate intensities and avoid fatigue-related injuries. The phenomenon fatigue can cause adaptations in the kinematic chain, but does not necessarily lead to a rapid loss in performance. For example, Saziorski *et al.* (1987) outlined a fitting model of three phases for short- to middle-distance runs. During the whole run, elite runners should have a deviation of 3 % in speed. From the beginning to the end of phase one, the runner has a certain relation between stride length and rate. The transition into phase two occurs approximately at the first third of the overall time and is initiated by an unconscious change of the neuromuscular pattern—stride length decreases and stride rate compensates to the speed of phase one. In the final phase, the stride rate also decreases, and so does speed. Based on these premises, the assessment of these changes—potential adjustments—requires a theory to explain and interpret, and proper methods of the measurement, recognition and analysis.

Thus, deciding whether or not a change is a neuromuscular adjustment or is within a typical range gives prominence to the observation of variability. "Variability is inherit within all biological systems" (Danion *et al.*, 2003, p.69) and has become of great interest in the biomechanical community. In 2007 Bartlett raised the following question:

> Is movement variability important for sports biomechanists?

Consecutive repetitions of a movement differ from one another in a certain extent (Davids *et al.*, 2006; Bartlett, 2004; Piek, 1998; Kelso, 1995; Winter, 1984). The cyclic movements of a runner are not identically equal to each other, but similar. Traditionally, variability has been regarded as "the amount of present noise in the perceptual-motor system" (Piek, 1998, p.143). This cognitive school considers "variability as reducing with skill learning as the learner controls unwanted degrees of freedom in the kinetic chain" (Bartlett, 2004, p.523). For example, variability is smaller in elite runners than in amateurs or novices (Nakayama *et al.* (2010); Glazier *et al.* (2006, p.50). This traditional thinking pools some assumption of sports biomechanics i.e., motor invariance, optimal motor patterns or movement techniques and a hierarchical approach to motor control (Bartlett, 2004, p.521). A deeper look makes clear that movement variability "should not be considered as an operational measure but also as a phenomenon of theoretical interest in its own right" (Piek, 1998, p.141).

The complex behavior of physiologic systems defy these traditional approaches (Goldberger, 2002), which furthermore assume regulation "according to the classical principle of homeostasis whereby physiological systems operate to reduce variability and achieve an equilibrium-like state" (Billat *et al.*, 2003, p.28). On the contrary, a growing body of literature argues that time series of physiologic signals contains "hidden information" (Goldberger, 2002, p.2466). Variability exhibiting in a special manner—far away from noise—emerges also, or especially, under resting, fresh and preferred conditions in nearly all physiologic systems. For example, the resting heart beat is known for its erratic character (Goldberger, 2002) likewise the stride time of walking at preferred speed (Jordan *et al.*, 2009, 2007b), see fig. 1.4. Variability contributes to the natural strategy—supporting the ability to adapt to changes in the environment (Jordan *et al.*, 2007b; Davids *et al.*, 2006), and hence, has functional meaning (Bartlett *et al.*, 2007). Thus movement variability might be important by providing insights into motor control

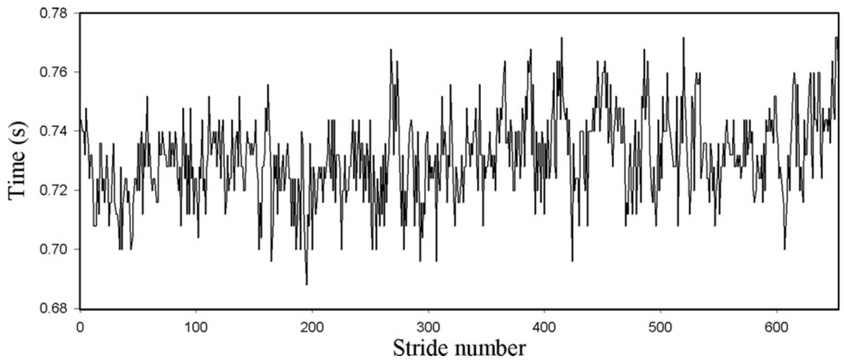


Figure 1.4: Stride interval time series for running at the preferred running speed. Although the coefficient of variation is quit small, the fluctuations are not random but have an erratic nature. (Jordan *et al.*, 2007a, p92).

and has the potential to support an assessment within an MC system.

In walking and running, the coefficient of variation (CV, mean divided by the standard deviation, STD) ranges from 1 to 5 % (Jordan *et al.*, 2007a,b, 2006; Cottin *et al.*, 2002; Hausdorff *et al.*, 2001)—this is a relatively small amount. In fact, "statistical measures of variability do not quantify how the locomotor system responds to perturbation and are not correlated with measures of local dynamic stability. [...] Local instability does not substantially adversely affect the [global stability of a cyclic movement]" (Dingwell & Kang, 2007, p.586), even though a certain amount of variability works to support global stability (Srinivasan & Mathiassen, 2012). The structure of stride-to-stride variability—considered as fluctuations—rather than its degree became of interest (Piek, 1998, p.141). In the last decade, it has become apparent that the fluctuations of temporal and spatial gait variables in both walking and running exhibit a nonlinear behavior being independent from the average stride (Hausdorff, 2007). Ever since, a variety of nonlinear dynamical approaches to motor control has been evolving with working elucidating and quantifying the terms *stability* and *variability* and their relationship with each other. These approaches account for the attributes of biological systems i.a., openness, non-linearity, self-organization, autopoiesis, and so forth, and are related to dynamical system theories such as the chaos-theory, synergetics and the gestalt theory (Witte, 2002).

As early as in 1995, Hausdorff *et al.* reports that the human walking gait exhibits "long-range [statistically] self-similar correlations extending over a hundred of steps", "as seen in a wide class of scale-free phenomena" (Hausdorff *et al.*, 2001). Scale-invariance and self-similarity are properties of fractals. Indeed, physiologic time series can be also ascribed to fractal-like phenomena ($1/f^{\beta}$) (Hardstone *et al.*, 2012; Torre & Wagenmakers, 2009; West & Latka, 2005; Goldberger, 2002; Hausdorff *et al.*, 1995). "The dependence of stride intervals decays in a power law, fractal-like manner with time" (Jordan *et al.*, 2006, p.120). Fractal-like behavior as well as long-range and short-range correlation in human walking and running change with age, health, fitness, cognitive load and speed (Bollens *et al.*, 2012; Jordan *et al.*, 2009, 2007a,b; England & Granata, 2007; Hausdorff, 2007; Jordan *et al.*, 2006; Beauchet *et al.*, 2005; Goldberger, 2002). Especially, alteration in these gait dynamics have been associated with the determination of disease severity. These approaches allow for displaying risks or presence of higher level gait disturbance, fall, parkinson's and huntington's diseases. Nevertheless, stability and variability reflect aspects of motor control and can be partly quantified.

The fractal dynamics and the auto-correlation have been commonly determined by a mono-fractal method, called Detrended Fluctuation Analysis (*DFA*). This method is a "classic root-mean square analysis of a random walk" (Terrier & Deriaz, 2011, p.4) and its solution is the scaling exponent and self-similarity parameter α, which indicates the correlation of a feature in a time series. Unlike *DFA*, the power spectral analysis is sensitive to nonstationarities and noise (Hausdorff *et al.*, 1995). It elicits the scaling exponent β. Both exponents can be transformed into each other. A method to quantify the local dynamic stability is the determination of the maximal Lyapunov exponent (λ). Local dynamic stability "refers to the sensitivity of the system to infinitesimal perturbations" whereas the "global stability refers to the ability of the system to accommodate finite perturbations" (Dingwell *et al.*, 2001, p.27). The global stability can be estimated with the help of the maximum Floquet multipliers ($MaxFM$) (Dingwell & Kang, 2007). The $MaxFM$ assumes a periodic system, therefore, the global stability is represented by the orbital dynamic stability.

All of these methods have been successfully applied to gait analysis, first to walking and then to treadmill running. Most of the studies inspected temporal parameters i.e., the stride time. The basic message is that the relation between the dependent variables and the factors can be expressed as a U-shaped function, whereby the y-axis denotes the scaling factor and the x-axis the speed, age, health, fatigue or cognitive load. The minimum of this dependent variable defines a scaling factor which expresses the stability of the motion. A scaling factor in the range between 0.5 and 1.0 denotes a positive correlation between successive movements. A typical value around 0.8 generally denotes stable movement. A persistent correlation i.e., stable movements, depends among others on the preferred speed and stride pattern, a young age, healthy, no fatigue, and little cognitive load. Values outside of the range from 0.5 to 1.0 indicate loss of stability in the repetitive movement, with lower values approaching white noise (0.5) or brown noise (1.5). A variety of modifications of these methods fitting the needs of the circumstances in each case can be selected, even more related methods e.g., approximate entropy and multiscale entropy analysis. But for all that gait analysis gains a lot from methods originated in other branches of scientific inquiry such as robotics, physics or even pure mathematics. In contrast to classic statistical analysis, these methods portray underlying principles of nature and are promising in the assessment of the human gait of both walking and running. This thesis aimed to apply these methods to make inroads in the assessment of over-ground middle-distance running.

How might this knowledge be exploited within an MC system? Having the U-shaped relationship in mind one would guess that there is a change in the pattern of biomechanical parameters during the course of a full-exhaustive run. Of course, the resulting movement, especially in advanced runners will be quite stable (with respect to the analysis of the degree and maybe as well as in the structure of variability) over a long time. Adaptations occur in the kinetic change spread over the whole range of degrees of freedom. The famous experiment by Arutyunyan and colleges (1969; 1968) of a professional pistol shooter demonstrated that an advanced but fatigued athlete can find compensatatory strategies in a way that the resulting movement remains stable. In this meaning, it is challenging to find alterations in features picked from the resulting movement, although expecting persistent kinematic variability. However, examination of the literature revealed that, at least in the analysis of the walking and treadmill running meaningful conclusion could be drawn upon gait properties.

The findings of Cottin *et al.* (2002) are of note—"velocity variability did not increase with fatigue" over the course of middle-distance all-out runs—neither in constant-pace nor in free-pace conditions, even though the oxygen kinetic was not affected. Thus, speed on its own does not seem to be a sufficient criteria within an MC system. In theory a constant speed ensures the most economical running strategy. However, Cottin and colleges summed the extra cost due to free-pace running up to lower than 1 % of total cost. The speed of a stride is the product of stride length by stride frequency. Both relate to each other but "have distinct effects on gait variability" (Danion *et al.*, 2003, p.76). It is of speculation—the adjustment of both attempts to drive the system towards a multiple of its own reasoning frequency, "for which the amount of energy needed to sustain its oscillation is minimal (Danion *et al.*, 2003, p.75)". The computer simulation of Miller *et al.* (2012) endorsed the minimum energy hypothesis. Cavagna *et al.* (1997) concluded that "at low-to-moderate running speed, the body operates at it's resonant frequency." However, the third last mentioned study proved that preferred running speeds are not necessarily optimal in terms of spatial and temporal variability i.e., energy expenditure is not minimal due to not reaching a harmonic of the reasoning frequency. Nonetheless, "long-range correlations are reduced at the preferred running speeds" (Jordan *et al.*, 2007a, p.88). Obviously, $1/f^{\beta}$-noise is not suitable for estimating the efficiency. Then again, it has been appreciated that $1/f^{\beta}$-noise changes with the speed in a U-shaped manner, but the influence of fatigue has not yet thoroughly been studied. It was a main concern of this thesis to find whether or not an MC system would benefit from drawing upon the development of the introduced indices concerning stability and variability.

Meardon *et al.* (2011) have provided valuable incentive to enforce continuing along this front. An intriguing outcome of their work was that the "distributional measures of variability did not increase over the course of the run [...but...] long-range correlations decreased over the course of the run" (Meardon *et al.*, 2011, p.38). They used *DFA* method. The subjects were instructed to run at their preferred 5 km pace (± 5 %) until cessation induced by fatigue. The group with history of injuries "demonstrated lower low range correlations" (p.38). As expected, fatigue is related to a loss of stability.

According to Bonacci *et al.* (2009), 50 % to 75 % of the injuries in running stem from overuse. Overuse injuries diminish the quality of training or lead to restriction in training. Both options convey the risk of loss in performance. "Alteration in neuromuscular control—imbalances, altered muscle timing, muscle fatigue and muscle weakness—have been associated with musculoskeletal injury and pain" Bonacci *et al.* (2009, p.915). If long-range correlations have the strength to indicate overuse or the training zones with preferred stability, then an MC system could supervise the runner. In particular, the topic 'intervention and feedback' may come to different insights than in the past. Incalza (2007) stated that it is disadvantageous to interfere with the natural movement pattern of runners and "hint that an individual athelete's running technique cannot be effectively modified", and, referring to Cavanagh & Williams (1982) Incalza, continued "any attempt at modifying the stride length or stride frequency produces a negative effect on the mechanical efficiency" (p.42). Nevertheless, the diagnosis of critical and sensitive phases may be helpful in reducing the risk of overuse.

To the knowledge of the author, only one study investigating the fractal structure of variability in over-ground running has been conducted—by Meardon *et al.* (2011). A uni-axial low-mass accelerometer attached to the tibia measured the stride impact. The time series of the stride time was determined by filtering and finally detection of the local maximum peaks. No study concerning over-ground running has yet been conducted investigating the stride length or any combination of length, cadence, flight time and contact time. However, in the context of PvC, the assessment of walking gait has reached a clinical level—a huge variety of measurement equipment and analysis techniques are offered to clinicians. In this light, accelerometer-based systems enable small, light weight, portable, hindering free measurements, attachments nearly anywhere on the body, the determination of temporal and spatial parameters, and so on. If necessary the determinations can be improved by aggregating accelerometry sensors, gyroscopes and magnet sensors to inertial measurement units (IMU). For example, Liu *et al.* (2009) introduced a wearable sensor system which was "designed to detect gait phases including initial contact, loading response, mid-stance, terminal stance, pre-swing, initial swing, mid-swing and terminal swing" (p.978) with the help of some IMU, also in 2009 Takeda *et al.* published a similar system. Attaching only one IMU to the pelvis, Koese *et al.* (2012) extracted the bilateral step length. However, Global Position Systems (GPS) have been applied for this purpose, too. For example, in subsequent studies of the years 2000, 2000, 2003, 2005, and 2005 groups of Terrier, Ladetto, Schutz, Merminod, and Gabaglio studied the biomechanics of human locomotion, especially walking gait. Proficient to their work was the determination of the step parameters i.e., length, cadence, time, with the use of a differential GPS (DGPS) worn in a backpack. They also examined the variability of the data—then in the year 2005 *DFA* method elicited long-range correlations in the fluctuations. In running, however, IMU-based studies (Yang *et al.*, 2011; Neville *et al.*, 2011) could be found determining the over-ground velocity, and identifying symmetry in the gait (Lee *et al.*, 2010b).

There is one more approach of interest that integrates both GPS and IMU—called Pedestrian Deadreckoning (*PDR*). *PDR* determines the present or a future position of a navigator from a known past position and the steered course and its velocity. Here, a fusion algorithms carries for the efficient balancing of each sensor types' advantages and drawbacks. As previously mentioned, due to the higher velocities and shorter contact times, running measurement has higher requirements of sensors and computational power. Among a variety of *PDR* approaches such as Jimnez *et al.* (2011), Tan *et al.* (2008) and Ladetto *et al.* (2001), there was one attracting to this thesis—Fischer *et al.* (2012). They published a detailed tutorial for *PDR* of walking and shared their research how to use foot mounted inertial sensors, in order to determine the position of a pedestrian. Instead of using only one foot, both feet were equipped with inertial sensors, in order to detect each step. In contrast to most *PDR* approaches, there was no typical heading information but the knowledge was applied that the athlete runs in a lane of an athletic track.

The thesis started when the MC system was at a very early stage. The beginning was coined by a lot of hardware issues and the establishment of the infrastructure between the server and the client. In the following time, sensor selection and integration struggled the work. Gradually, the author became aware of the immense meaning of the analysis and diagnose—if a biomechanical grounded MC system should ever be used in a running training session. Regarding the various studies confirming that a biomechanical assessment in running may capitalize on fractal analysis methods, thus time series of stride parameters were seen as a requirement to this work. On this as a basis, the thesis attempts to contribute to some of the introduced open questions towards running.

1.2 Aims

The thesis is a report displaying an excerpt of the development towards a biomechanically based MC system. The primary aim of this thesis was to establish, test and validate a prototype for MC for running with the introduced features. Herein, the feedback management has a crucial function. The subordinated aim was to explore its merits and limitations. As a particular feature, this MC system should be available as a low-cost system using common smart-phones and commercial sensors.

The second part is dedicated to the aim of establishing a time series of selected biomechanical stride and step parameters. The *PDR* algorithm of (Fischer *et al.*, 2012) was adapted to the requirements of running, in a way that for consecutive strides, the following parameters could be determined: length, cadence, contact time, flight time and speed. For further processing, these parameters were validated in a running scenario.

The ensuing goal of the thesis addresses the issues of stability and variability in a full exhaustive middle-distance run observed by an MC system. The development of the stride parameters over the course of the run are illustrated. The overall questions were: 1) Do running kinematic characteristics change over a 5 km run? 2) Can the run be divided into phases e.g., as proposed with the model introduced by Saziorski *et al.* (1987)? 3) How do biomechanical, anthropometric and physiologic variables relate to each other, and to speed and fatigue? 3) How do indices of stability and variability behave? 4) Which conclusions can be drawn for performance and the risk of injury? 5) What recommendations can be given for an MC system, especially regarding feedback during running?

Finally, the thesis called for an answer to the questions, how can the gained knowledge be applied to the prototype. The *PDR* approch comprises two 3-d IMU, whereas the prototype of the MC system for runners makes use of one accelerator sensor. Is it foreseeable that such a minimal configuration can handle the requirements of intervention and feedback management?

The aims of the thesis are listed in short below:

1. Establish a prototype for MC.
2. Adapt and evaluate the *PDR* algorithm (Fischer *et al.*, 2012).
3. Perform a biomechanical assessment related to the aformentioned issues, in order to prepare for feedback management.

1.3 Method

The author undertook a comprehensive literature research, in order to examine the above-mentioned issues and design a project plan. Our team worked on the realization of an MC system. In collaboration, we developed strategies and solutions for the topics: hardware, software, server, client, sensors and feedback routines. For the sake of the evaluation of the prototype, Veronika Kremser supported the author with her master's thesis (Kremser, 2011). The *PDR* solution was adapted with the help of Georg Ogris. Measurements of the IMU and GPS were fused (IMU/GPS) regarding the constraint that a runner is on one lane of the athletic track during the entire run. Different strategies were tested and compared. A physiologic performance diagnostic for a $5\,km$ run under full exhaustion was conducted as the final experiment to test the prototype and the *PDR* solution, and further to explore stability and variability in the gait of running. The reference system consisted of a combination of a radar measurement device and IMUs (IMU/RADAR). As a note, in the beginning of the development of the prototype, one goal was to integrate high resolution analog inertial measurement units (at least acceleration sensors) to determine stride parameters such as the stride length with sufficient accuracy and precision. Therefore, a micro-controller was planned to assist as an interface receiving analog signals and converting them to a digital stream of the incorporated BSN. Within the time frame, this goal could not be achieved, therefore, an alternative solution made use of a separate measurement equipment for the *PDR* approach.

1.4 Guidance through the book

The following chapter 'Related work' contains the literature review. The first section offers a background to the topic 'Performance of distance running' to the reader. Due to the purpose of the thesis the section 'biomechanics of running' and its related topics have a centrality between the distinct factors on performance. These sections mainly broach the issues of lower limb movement, the phases of the cyclic movement and the relationship of stride and step parameters to each other and to velocity and how this is influenced by anthropometry, training and equipment. Section 2.1.7 reveals the biomechanical adjustments over time of human running. In particular, it examines controversy between existing studies. The section 'Managing stability and variability', starting on page 78, is of prime importance. The concepts of stability and variability in human movement are introduced, also their analysis methods are discussed.

Chapter 3 outlines in short the issues for the research and finishes with certain questions and hypotheses for the (statistical) analysis.

In the middle of the thesis, the architecture and implementation of the prototype of an MC system for running is illustrated. Herein, the reader will also find a description of the feedback provision. The feasibility of the prototype for running scenarios was examined by evaluating kinematic parameters. In the ensuing chapter, the author describes the adaptation of the *PDR* algorithm from page 113. The fusion algorithm has a focal point, which finally leads to the determination of stride parameters based on IMU/GPS.

The experiment of the middle-distance run regarding stability and variability and its analysis by the reference system is presented in chapter 6. Thereafter, the *IMU/PDR* method is applied to the 5 km experiment. The results are compared to those of the reference system and prototype. At the end of this chapter on page 153, a discussion is held about the transformation and implementation on the MC system.

Summary and conclusion are given on page 165.

2

Related works

Performance is the main criteria both in elite running and in amateur and recreational running. Runners dispose of limited energy resources and intend to transform most of this amount into propulsion i.e., they attempt to cover a desired distance in minimal time by running as fast as possible. Performance can be influenced by the effects of fatigue. Particularly, muscle fatigue may lower overall performance, because the force-generating capacity of a muscle or muscle group has been reduced by activity (Gates & Dingwell, 2011). These authors defined fatigue "as an inability to sustain a target work rate“ (p.525). Fatigue under low intensities does not reduce the performance but increases the perceived effort. As a complex phenomenon fatigue "may also induce changes in the motor coordination“ (p.525). Some of those changes have been found to cause an athlete to "develop injuries by inducing poor biomechanics" (p.525). Fig. 2.1 gives an overview of the theoretical possibilities on the effects of fatigue. The following sections will have a look at the factors of performance and how biomechanical (kinematic) parameters or stride patterns may predict changes in performance. The risk of injury has been associated with the neuromuscular control system and its changed potential over the course of an intensive run. To illustrate this potential, computing methods were investigated to quantify stability and variability of cyclic movements. Furthermore, measurement methods used in outdoor running were compiled and evaluated for this purpose.

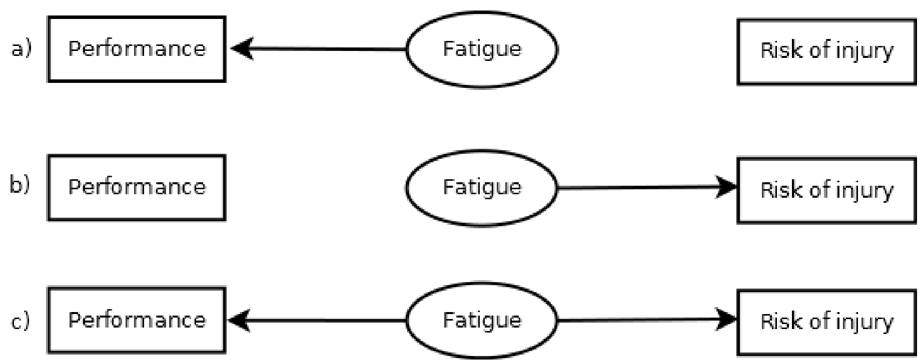


Figure 2.1: Effects of fatigue a) on performance b) on the risk of injury and c) on both

2.1 Performance

This section gives an overview of the factors on running performance to the reader. Performance is a key word in sports—athletes strive for it. An appropriate performance is required to achieve a desired goal. Athletes take measures to change their performance for the better. The athlete is exposed to disturbances and recreation. The interplay between these influences leads to changes in the athlete and thereby in overall performance. Improving performance is associated with risks of degeneration, too. It is a narrow ridge between effective load and over-stress. The higher the performance is to be achieved, the higher the risk of over-stress. However, the rate of injuries in recreational runners demonstrating lower absolute and relative performance is also significant (Novacheck, 1998; Bonacci *et al.*, 2009) for a variety of reasons. Physical condition and body awareness may be of importance to this. On the one hand, both professional and recreational runners aim at accomplishing a desired performance in training for health reasons or better results, and on the other hand, attempt to lower their risk of injury to a minimum. The ongoing subsections consider this matter from several angles, see fig. 2.2, with the biomechanical factor, most relevant to the purpose of the thesis, receiving the greatest attention. At the end of this section, the reader should be aware of the factors contributing to performance and when performance reaches a critical level. The overall intent was to demonstrate how an observation of the runner may support the two main goals of sports biomechanics—improving performance and reducing the risk of injury.

2.1.1 Economy

Performance in sport has a broad meaning and usually refers to a measure that allows for assessing the athlete's action. Strictly speaking, performance relates to the result of the task—in running it is horizontal velocity. In endurance disciplines, the performance is associated with the mechanical power (P), which can be defined as the rate at which energy (E) is consumed or positive work (W) is done, see eq. 2.1. The energy available to the athlete is limited by physiologic capacity, hence, economy is directly connected to performance (at least at sub-maximal speeds), see fig. 2.2. In this meaning, the cost of resources has to be minimized i.e., an optimal running economy is the minimal energy needed to travel a distance at a certain speed, or in other words, a runner requests one's

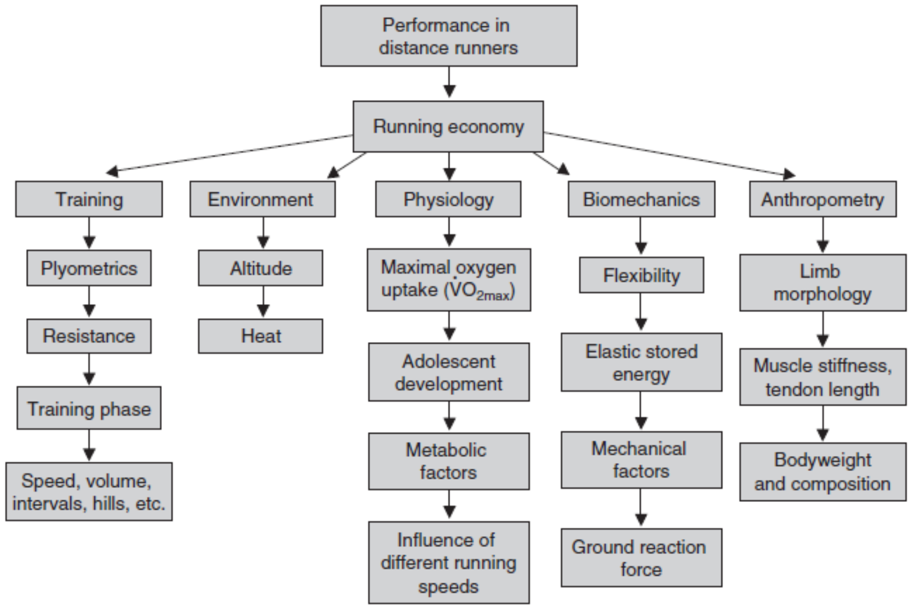


Figure 2.2: Factors affecting running economy (Saunders *et al.*, 2004)

maximal energy rate according to the desired distance with maximal transformation in horizontal propulsion. It will be shown that a variety of factors contribute to the complex structure of running economy and performance. Furthermore, to date, which and to what extent these factors describe the degree of its effectiveness and efficiency to running economy has remained elusive.

$$P = \frac{dW}{dt} \quad P \ldots \text{mechanical power}, \quad W \ldots \text{mechanical work} \tag{2.1}$$

This may arrive due to several reasons, which have not yet been seen fully elucidated by research. First, it has to be grasped the total energy yield, second, how much the relevant factors contribute to it, and third, the minimum credible amount of energy—the optimum. Economy can be expressed as the relationship between the physiologic demand and running speed, whereas mechanical efficiency (ME) "is the ratio between the mechanical energy produced during the exercise and the energy cost of the exercise“ (Karp, 2008, p.5802), see eq. 2.2 or the relation between the input and output of mechanical power (Robertson, 2004). Thus the mechanical demand and its determination will be elucidated in the following, whereas subsection 2.1.5 carries about the physiologic issues.

$$ME = 100\,\% \cdot \frac{W_{output}}{W_{input}} \tag{2.2}$$

$ME \ldots$ mechanical efficiency, $W_{output} \ldots$ also net W

Various studies have been undertaken attempting to determine the amount of mechanical work and power, see e.g., Nummela *et al.* (2007), Purkiss & Robertson (2003), Chang & Kram (1999), Roberts *et al.* (1998), Willems *et al.* (1995), Flynn & Soutas-Little (1993), Kram & Taylor (1990), Williams & Cavanagh (1983), Cavagna & Kaneko (1977), Fukunaga *et al.* (1980), Cavagna *et al.* (1977), Cavagna *et al.* (1964) and Williams & Cavanagh (1987). Purkiss & Robertson put emphasis on importance of the 'internal biomechanical cost', in order to "distinguish between efficient and inefficient motion“. The determination of mechanical work and power is challenging in many ways. For example, the 'zero work paradox' has led researchers to include a grade (an angle of incline) greater than zero in treadmills while studying human gait (Purkiss & Robertson, 2003). In human locomotion, the total mechanical energy consists of 1) gravitational potential energy, 2) elastic potential energy, 3) translational kinetic energy, 4) rotational kinetic energy, 5) energy transfer in segments, 6) energy transfer between segments, and 6) friction. When it is difficult to account for "elastic energy storage and release" (Sasaki *et al.*, 2009, p.739), it is usually omitted. This is one reason why, measures of mechanical power range widely (Williams & Cavanagh, 1983; Winter, 1979). A traditional approach is the calculation of the external mechanical work with respect to the rate of energy changes at the center of mass:

$$W = \int |\dot{E}_{CM}| \, \cdot \, dt \qquad E_{CM} = PE_{CM} + KE_{CM} \tag{2.3}$$

$CM \ldots$ center of mass, $PE_{CM} \ldots$ potential energy, $KE_{CM} \ldots$ kinetic energy

$$KE_{CM} = \frac{1}{2} \cdot m \cdot \vec{v}^{\,2}_{CM} \qquad PE_{CM} = m \cdot g \cdot h_{CM} \tag{2.4}$$

$m \ldots$ mass, $g \ldots$ gravity, $\vec{v}_{CM} \ldots$ velocity of CM, $h_{CM} \ldots$ vertical height of CM

The total work done by muscles is smaller than about four times of external work in the walking gait (Cavagna & Kaneko, 1977). "In running, the efficiency increases steadily with speeds (from 0.45 to 0.80) suggesting that positive work derives mainly from the passive recoil of muscle elastic elements and to a lesser extent from the active shortening of the contractile machinery" (Cavagna & Kaneko, 1977, p.467), so up to $20\,km\,h^{-1}$ $\dot{W}_{int} < \dot{W}_{ext}$ and at higher speeds, vice versa. Later Kaneko (1990) confirmed this result. According to Arampatzis *et al.* (2000), this approach showed a moderate relation between running speed and mechanical power, whereby this relation is assumed to be linear due to the linear relation between the rate of oxygen uptake as a measure of metabolic energy and speed. In contrast, the integration of the ground reaction forces (GRF) yielded a good relation to the mechanical power, see eq. 2.5. However, these external work approaches underestimate the total work, even when corrected by scales, as they do not provide insights into the kinetic chain or the understanding of related efficiency. In particular, the point mass model "neglects the large energies (translational and rotational kinetic) associated with the reciprocal movements of the legs and arms" (Winter, 1983, p.91).

$$W = \int |L_R| \cdot dt \qquad L_R = \overrightarrow{F_R} \cdot \overrightarrow{v_{CM}} \tag{2.5}$$

$$L_R \ldots \text{P of GRF}, \quad \overrightarrow{F_R} \ldots \text{GRF}, \quad \overrightarrow{v_{CM}} \ldots \text{velocity of CM by GRF}$$

Estimations including movements of body segments yield stronger relations between speed and mechanical power, but only if they make use of the GRF; pure kinematic approaches yield weak correlations as Arampatzis *et al.* (2000) stated (see also eq. 2.8). Based on this notion, research has focused on the estimation of the internal work permitting energy transfers such as within and between segments, friction, and elastic storage or release. For example, the inverse dynamics method, using GRF and kinematics, applies the Newton-Euler equations of motion from the most distal joint of the rigid body model on upwards. These equations "compute the powers produced by the moments of force at the joints then integrate them with respect to time to find the work done" (Arampatzis *et al.*, 2000, p.144). In general, these complex calculations require high computational effort. Nonetheless, these approaches consider the net mechanical work, which may have both a "backward component" and one for the "forward push". Both negative and positive work sum to the net mechanical work as they occur in the same phase (Fukunaga *et al.*, 1980; Winter, 1979), see fig. 2.3 and eq. 2.6. It also may be conceivable that "co-contraction of antagonist muscle groups and separate individ-

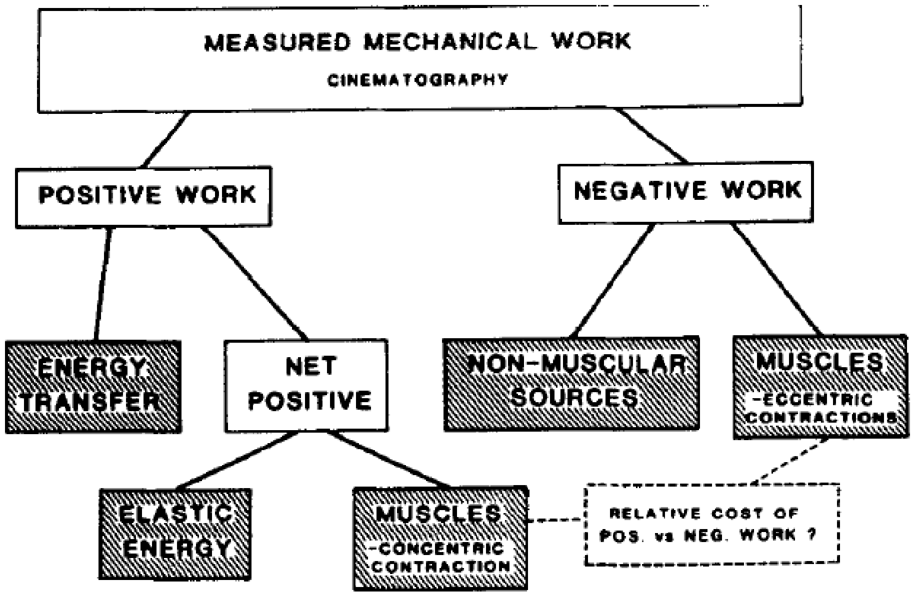


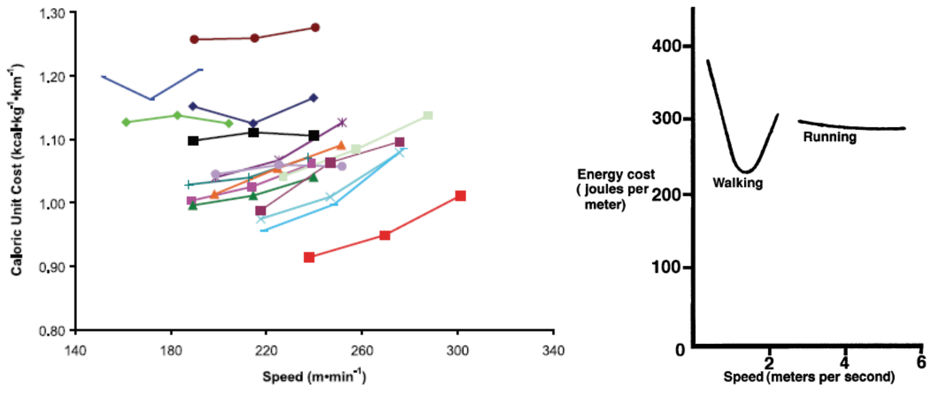
Figure 2.3: "A diagram depicting sources of work and factors which will influence the relationship between mechanical work derived from cinematography and the metabolic work associated with it" (Williams & Cavanagh, 1983, p.116).

ual muscle fiber and tendons" (Sasaki & Neptune, 2006, p.384) produce not negligible work due to an uneconomic running technique or fatigue. Pierrynowski *et al.* (1980) demonstrated that "normal walking is a highly conservative movement with about two-thirds of the segment energy changes being due to passive exchanges with segments or between adjacent segments. Only about one-third of the energy change, called internal work, can be attributed to muscular work."

$$ME = 100\,\% \cdot \frac{W_{external} + W_{internal}}{W_{input}} \qquad ME \ldots \text{mechanical efficiency} \tag{2.6}$$

$$W_{tot} = |W_{external}| + |W_{internal}| \qquad W_{tot} \ldots \text{total work} \tag{2.7}$$

Willems *et al.* (1995) emphasized the importance and the problems determining the internal work. The internal work is referred to the work of all the acting muscles, whereas the external work is responsible for the replacement relative to the ground. External and internal work sum up the net mechanical work, but the total work may



(a) (b)

Figure 2.4: (a) Caloric unit cost Fletcher *et al.* (2009, p.1921)(b) O_2 curve Novacheck (1998, p.90).

be greater than the net work, see eq. 2.7. These investigators example the "possible transfer of energy between internal and external work components of the total work" (p.380). This, once again, demonstrates the intricacy involved in estimating influences of different biomechnical factors on running efficiency. Therefore, the next example of mechanical work is the approach regarding the movement of the body segments, see eq. 2.8. Herein, Winter (1979) calculated the average error to 16.2 %, "but can be as much as 40 %" (p.82). However, Arampatzis *et al.* (2000) evaluated this method having poor predictivity between mechanical power and speed.

$$E_b(t) = \sum_{i=1}^{n} PE(i,t) + \sum_{i=1}^{n} KE(i,t) + \sum_{i=1}^{n} RE(i,t) \tag{2.8}$$

$b \ldots$ body, $n \ldots$ number of segments

$i \ldots i$th segment at time t, $RE \ldots$ rotational energy

Running is more efficient than walking at speeds above the preferred walk-run transition (Sasaki & Neptune, 2006), even if the influence of muscles in running is greater (higher internal work) compared to walking. The energy cost for walking can be represented by an U-shaped function (Umberger & Martin, 2007), see fig. 2.4, b. It is minimal at the preferred walking speed—under or above that, the energy cost (per unit distance traveled) increases. Numerous studies, among them Cavagna *et al.* (1964), proposed that "the external work per kilometer is independent of speed, amounting $0.25\,kcal\,kg^{-1}\,km^{-1}$" (p.249). According to Cavagna *et al.* (1977), there are two basic mechanism for minimizing energy expenditure in bipedal terrestrial locomotion. The first one can be referred to walking and concerns a pendulum model, which accounts for the "exchange between gravitational and kinetic energy." (p.R243). The second one accounts for the recovery of mechanical energy stored in muscles as "both kinetic and gravitational energy". The elastic structures cause a bouncing effect—a recoil. Fig. 2.4, b depicts that running "at top speed or leisure pace" is nearly the same in meaning of energy cost (Nummela *et al.* (2007, p.656), (Lohman *et al.*, 2011), Roberts *et al.* (1998), Kram & Taylor (1990), Martin & Morgan (1992)). These studies argued that the main energy is spent on lifting and accelerating the body and limbs. The propulsion has a relatively minor influence on energy expenditure. At speeds greater than $2.2\,m\,s^{-1}$, running efficiency reaches a plateau while vertical displacement is minimal and the bouncing effect maximal. Fast runners are economical—their metabolic rate is smaller in comparison to similar runners at given speeds, see fig. 2.4, a. Roberts *et al.* (1998) also supports the hypothesis "that it is primarily the cost of supporting the animal's weight and the time course of generating this force that determines the cost of running". In the study of Fukunaga *et al.* (1980), the power in forward direction (P_f) at speeds lower than $5\,m\,s^{-1}$ was always less than the power against gravity (P_v). At speeds higher than $6\,m\,s^{-1}$, P_f increased gradually greater than P_v. As a function of speed, P_v increased linearly, and P_f as the 2nd power of running speed, see eq. 2.9 and 2.10, respectively. Taken all together, this hypothesis says that running is efficient at nearly all (common) speeds—that is, cost is speed invariant.

$$P_v = 0.878 \cdot v_f + 8.55 \tag{2.9}$$

$$P_f = 0.436 \cdot v_f^{2.01} \tag{2.10}$$

$P_v \ldots$ P against gravity, $P_f \ldots$ P in forward direction,
$v_f \ldots$ velocity in forward direction

Steudel-Numbers & Wall-Scheffler (2009) claimed that there is also an optimal speed in running. In contrast to previous findings, this study did not fit the data points by a linear regression model, but rather used a quadratic model with a higher correlation coefficient ($R^2 = 0,984$). They also pointed out that "the majority of earlier work on human running looked at the cost per unit time [...], rather than the cost per unit distance" (p.356). Nonetheless, this "small difference" (p.357) revealed a weak curvilinear relation between speed and energy cost and comes as a proper puzzle for the next considerations on economy. Chang & Kram (1999) examined the horizontal force and their contribution to the total metabolic cost of running. They showed that the "muscular force to support body weight is a major determinant" (p.1657) and quantified this relationship. More than one-third of metabolic cost is from horizontal muscle force production, and thus is not negligible. For example, windy situation have significant impact on metabolic cost. These findings are in conflict with the previous mentioned findings. Therefore, they suggested not to consider horizontal and vertical forces independently as "physiological constructs" (p.1661), but how these forces contribute to the net force. Either they are right, or this influence is lower; Roberts *et al.* (1998) lends support to the argument that the majority ($70 - 90\,\%$) of the increase in energy cost with speed is related to a small but consistent increase in the rate of force generation, or the decrease in "time available to generate force" (p.2745), respectively. Higher speeds rely on faster muscle fiber types which are "metabolically more expensive" (p.2745), for details see section 2.1.5. The time course generating force but not the ground reaction forces itself is a determinant of economy. Long contact times and significant deceleration is seen to be wasteful (Kram & Taylor, 1990). This does not necessarily mean that the contact time (CT) should be minimized rigorously but emphasizes the appropriate loading of elastic energy storage. Martin & Morgan (1992) expresses this dynamic by stating "muscular effort appear to have the greatest potential for explaining metabolic energy demands during walking and running" (p.467) but admitted that "it is unclear what quantifiable descriptor can best reflect muscle force production" (p.467) and continues with the observation that ground reaction forces and economy have a weak to moderate relationship.

Next to short contact times, for example, lower medial peaks and lower first peaks were related to economy. The more a runner 'bounces' by mid to forefoot striking, the more the runner "rely [...] heavily on the musculature to assist with cushioning than rearfoot strikers" (p.471). This comes with Weyand *et al.* (2000). Their study stepped on this and put weight on this framework. If the elastic recoil is exploited, then ground reaction force does play the most important role but "not more rapid movement" (p.1991). During level treadmill running, the faster runners showed significantly greater mass-specific effective forces than the slower runners at their top speeds, see next section for details. Snyder & Farley (2011) experienced that stride rates greater than the preferred cause higher metabolic cost. It is still an element of debate whether or not, and why, there are multiple optimal stride rates, perhaps, due to muscle activation patterns. Some researchers suggested approx. "$2.7\,Hz$" (Cavagna *et al.*, 1997, p.2089) as the optimal step rate. This or these rates (frequencies) might be a compromise "between the minimization of mechanical work and maximization of elastic energy use" (Cavagna *et al.*, 1997, p.2089). Daniels (2005) observed in elite runners across middle and long distances that the best runners kept to a constant stride rate (SR) of $1.5\,Hz$. It has been suggested that the whole body of the runner can be considered a single linear spring. As the SR increases, the leg stiffness (and tendon hysteresis) increases, too, and thus enables an efficient recoil. However, the often cited U-shape function of running economy in dependence from the degree of the preferred SR remained stable for level, uphill and downhill running i.e., the energy cost increased below and above the preferred SR (Snyder & Farley, 2011). The computer simulations of Miller *et al.* (2012) "indicate that a variety of variables can be minimized to incur a realistic low [cost of transport]" (p.1503). Their study suggest that minimizing this cost does not lead to a realistic simulation of human running, but hinted that minimizing muscle activation and better timings around hip, knee and ankle joints could reduce metabolic cost. More freedom of motion with the link to optimization was found in the swing phase, whereas the stance phase is more mechanically constrained. It is not conclusively shown to what extent the swinging leg contributes to the cost. Some studies suggest a portion of up to 30 % of the total cost. A strong flexing of the leg may reduce the moment in the hip joints due to a shorter pendulum (reduced inertia) and in consequence prepare the activation of the leg extensors.

The proposition of Williams & Cavanagh (1987)—emphasizing "the importance of biomechanical influence on running economy" has been proven again and again. Some variables have been found responsible to a "substantial portion of variation in economy" (Morgan *et al.*, 1989), but there is no exclusive biomechanical parameter or subset of variables (Kyroelaeinen *et al.*, 2001; Williams & Cavanagh, 1987). In the context of this thesis, it should be mentioned that "running economy is affected by [...] stride length, [and] change in speed during ground contact phase [...]“ (Nummela *et al.*, 2007, p.655). The next section elucidates the biomechanics regarding those parameters.

2.1.2 Biomechanics

This section intends to make available the basic biomechanics regarding the focus of this thesis. For further details the reader is referred to Novacheck (1998), Ounpuu (1994), Adelaar (1986) and Dillman (1975). Technical terms were defined as they were used in the experimental part of the thesis. In the previous section, the stride and step parameters of interested were mentioned: speed, rate, time, ground contact time, flight time and length. Their relation to each other and relevance to the assessment in running are summarized in the following.

The approaches to running seen in science and in the consumer market reflect the aspiration to performance and prevention of injury. Especially since the 1970s, with the invention of the first running shoe, and improvements in athletic track composition, the view and development of running techniques and equipment have changed dramatically. Training regimes arousing the current debate about what human running gait is optimal. There is a strong believe that efficient biomechanics will be adjusted by high volumes of training automatically and that ”there is no commonly accepted running model which will suit everyone“ (Nytro, 1987)(cited by Romanov & Fletcher (2007)). In this line, Wallack (2004) postulated ”Form is God-given. If you systematize it, you destroy it”. Various studies point in this direction, too, thinking of the preferred stride rate and length, which both were near at the economic target. For example, often cited pioneers in this area like Cavanagh and Williams reported that interfering with these parameters would cause negative effects. The runners in their studies were measured to be nearly at their optimum—of their current configuration. The previous section hinted at biomechanical factors of economy. Since it has not been fully substantiated what determines running economy, these measures might represent only one optimum—regarding the current configuration. Several running methods have been developed—claiming to be economical, safe or both. An example of different techniques or variation, respectively, will be given later on. However, running is a natural movement. Early man ran to capture prey (Lohman *et al.*, 2011; Steudel-Numbers & Wall-Scheffler, 2009). Hunting had to be efficient due to long distances, fighting with

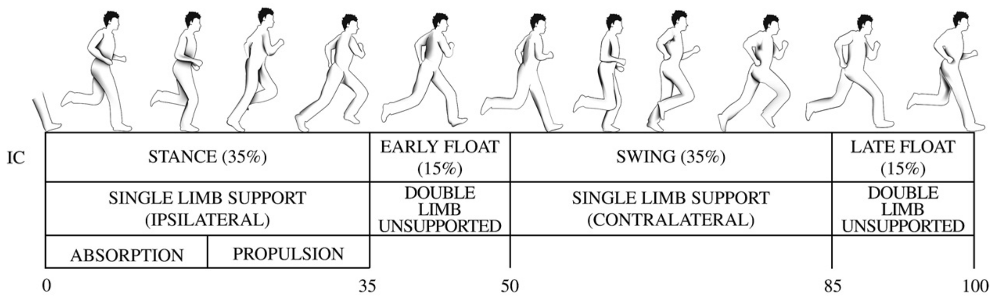


Figure 2.5: Running phases (Lohman *et al.*, 2011)

the animal and the transport of prey back to the camp. Since that time various factors impacting the running techniques have changed, but the need for learning to run remained. As a consequence, running technique has been adapted to these changes. For example, heel striking occurs owing to the cushioning of the shoe. Children imitate 'joggers' who mostly have non-natural technique, and so forth. Human and animals obviously differ by the motor control mechanism. While animals coordinate gait chiefly through spinal cord level signaling, humans' control mechanism involve higher level brain control (Duysens & Van de Crommert, 1998). There seems to be more room for adaptation. Nonetheless, a "thorough understanding of running gait" is regarded as necessary "to properly treat and potentially prevent running injuries" (Lohman *et al.*, 2011, p.161). The large volume of literature suggests that modern technology can support assessment in running—to the belief of the author as well.

Running is a cyclic movement, which can be divided into phases, see fig. 2.5. A stride cycle begins with the initial contact of one leg in front of the center of mass in direction of motion. In this moment, the weight acceptance occurs. Dependent on the distance between the initial contact and the line from the center of mass perpendicular to the ground there can be a braking component (negative work) which transfers energy into the ground. Muscle contractions in the leg extensors or the skeletal systems compensate this impulse. In this early stance phase, the running strategy decides how the elastic recoil can be exploited. Therefore, it is named 'loading response'. When the center of mass travels across the line perpendicular to the initial contact, the runner is in the midstance. From this time point on, the propulsion (positive work) has been initiated and lasts during the terminal stance until the toe off event. After the rear leg has left

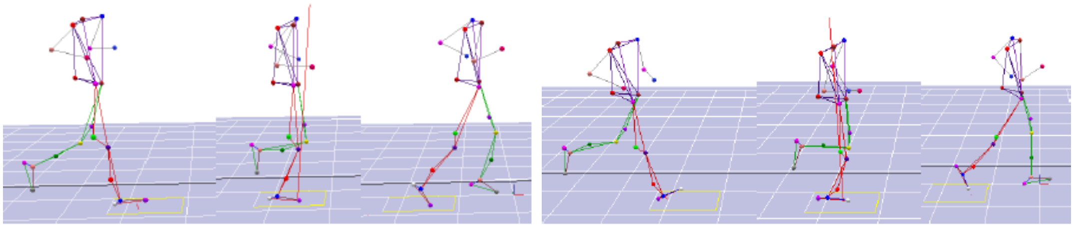
Table 2.1: Stride parameters

Parameter	Description
Stride time, ST	The time between two consecutive initial contacts of one leg.
Contact time, CT	The time between the initial contact and the toe-off of one foot.
Flight time, FT	The time while one foot has no contact to the ground.
Double float time	Both feet have no contact with the ground.
Stride length, SL	The distance between two consecutive initial contacts of one leg.
Stride rate, SR	Inverse of ST. Also called cadence or frequency.
Stride speed, SS	Average speed of the CM or foot during one stride.

the ground the early float has begun. Until the midswing, both legs do not support the body weight. This is called double float or airborne and ends up with the initial contact of the contralateral leg; the step cycle has terminated. At the midswing, both knees are in line; the contralateral leg is in the midstance. Finally, the terminal swing ends with the initial contact, as ends the stride and the second step cycle; eventually the stride cycle can repeat anew. The stride parameters used in most of the mentioned studies and with importance to this thesis are enclosed in table 2.1.

One of the reasons why it is beneficial to get deeper insights into the mechanics of running is a basic claim of research—to promote understanding. For example, Romanov & Fletcher (2007) developed a model of running that contradicts the current school of thought. In this model, gravity has centrality and is not something runners struggles with but the motive force that enables locomotion. In the last section, gravity was seen as a determinant in running, consuming most of the energy for lifting and accelerating the center of mass. It is reasoned to emphasize the push-off to better exploit the energy return, and thus propel the body forward. To further support this, the swing leg should be active to strengthen the catapult mechanism (Miller *et al.*, 2012); in early to midswing the lower leg should be flexed and kicked backwards; finally, in the mid to terminal swing it might be desirable to lift the knee to drag the body forward. The running model described by Romanov & Fletcher (2007) is employed here in place of the group of alternative methods in the next paragraph.

In comparison to a flic-flac or a height-jump, running does not expose an increase of vertical ground reaction force, see fig. 2.6 and 2.7. Sprint runners produce "relatively



(a) (b)

Figure 2.6: a) Pose running b) Heel strike running; accord. to Fletcher *et al.* (2010)

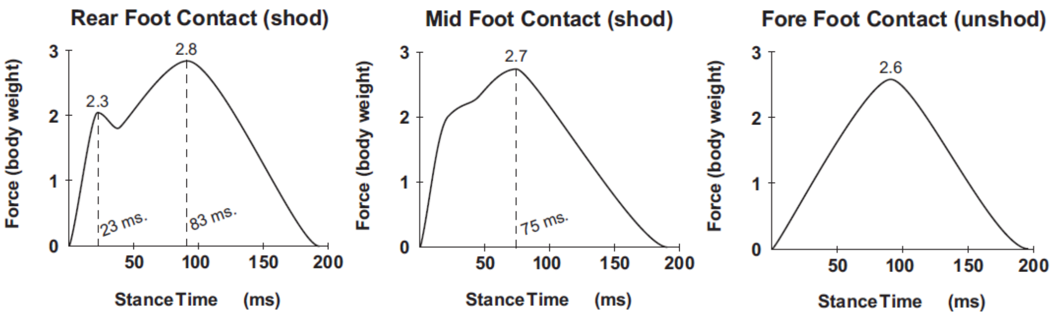
moderate vertical impulses“ (p.437). Keller *et al.* (1996) found that vertical ground reaction force increases linearly from 1.2 to 2.5 times body weight up to 60 % of the maximal speed, then remaining constant as runners i.e., sprinters lean forwards. However, the maximum of the vertical ground reaction force represents the end of weight acceptance, whereas afterwards the center of mass begins to rise. Moreover, ”a large vertically directed impulse cannot produce a horizontal displacement“ (p.437). During the mid to terminal stance, the extensor muscles are silent and this phenomenon became known as the ’extensor paradox’. A runner does not accelerate against the ground by pushing. Running can be compared to the passive walking model, which is used in robotics. ”The body falls forwards via a gravitational torque“ (p.441). The swing leg catches the body; prevents the body from falling to the ground. The closer the foot’s impact is at the projection of the center of mass to the ground, the less the braking component is. While the leg begins to support the weight, the quadriceps and the gastrocnemius are eccentrically active. Gravity causes them to stretch. At least midfoot striking is necessary to activate the gastrocnemius. The higher the speed is, the more the forefoot strikes and the better the elastic energy can be efficiently returned. The gravitational torque as the resultant of gravity and ground reaction force increase as the center mass travels forwards. The runner acts like an inverted pendulum. While the net torque is dragging the body forwards but also downwards, a radial acceleration which has been caused by the ground reaction force, changes the direction of the center of mass forwardly; finally, the center of mass rises due to the pendulum-like geometry of the runner. Conclusions based on this model have been drawn towards an efficient running method and attempted to realize (Dallam *et al.*, 2005).

There is a vast literature concerning biomechanical factors and recommendations of running technique. As an example, the summary of Morgan *et al.* (1989) is presented in tab. 2.8. For the sake of completeness, some criteria regarding the gravitational running model, will be given now to demonstrate that research in running is essential to define diagnosis and analysis in running. It is well accepted among most methods that little vertical oscillation is of advantage, also that the contact with the ground should be done by the outer edge of the foot. As above-mentioned, the initial contact nearby the vertical line under the hip and mid to forefoot strike reduces the absorbing energy. Heel striking reduces the efficiency of the shortening-stretch cycle and prolongs the CT. Although when forefoot striking, the heel has to hit the ground, in order to allow for the loading response. The runner does not need to push off the ground. The extension of the hip supports the radial acceleration of the center of mass. The work of the hip decides how close the swing leg comes back and how high the knee will be lifted: the faster the run, the more both. The 'pawpack phenomenon' is here considered to be a waste of energy, which is made difficult in this model due to slightly forward leaning body posture and the occurrence of impact under the hip. Instead of this the foot should be pulled forward to prepare the next initial contact after the toe has left the ground and the body has entered flight phase. In this model, smaller than the average stride length (SL) accompanied by higher stride rates are preferred. Tab. 2.2 compares research variables between heel-toe and Pose running. Therefore, twenty male competitive runners ran at a speed of $3.35\,m\,s^{-1}$. To demonstrate the sensitivity of such adjustments in running the stride parameters and their effects in concert will now be considered.

In general, the product of SL and SR is the horizontal velocity of the runner, thus a variety of combinations lead to a certain velocity. These combinations influence the economy. There seem to be preferred and optimum adjustments. Both stride parameters are seen to aid the assessment in the field. "With such, coaches will be better equipped to identify the characteristics of their athletes and to assign the exercise that are most appropriate for improving performance“ (Incalza, 2007, p.43). Fletcher *et al.* (2010) observed that for coaches a clear kinematic profile would be more helpful than kinetic measures. Nonetheless, with an increase in speed both parameters increase (Fukunaga *et al.*, 1980). Between 3.5 and $6\,m\,s^{-1}$, SL is dominant (Weyand *et al.*, 2000;

Table 2.2: Pose vs heel-toe running accord. to Fletcher *et al.* (2010)

Parameters	Pose	Heel-toe	P-value
$CT\,[ms]$	0.21 ± 0.01	0.25 ± 0.01	0.002
l_{stance} of $CM\,[m]$ (see eq. 2.11)	0.722 ± 0.053	0.874 ± 0.064	0.001
Knee flexion stance $[rad]$	0.41 ± 0.01	0.58 ± 0.04	0.001
Knee extension stance $[rad]$	0.46 ± 0.2	0.69 ± 0.01	0.04
Knee flexion angular velocity from terminal stance until mid-swing $[rad\,s^{-1}]$	7.8 ± 1.0	6.1 ± 0.9	0.04
Shoulder, hip ankle vertical alignment at 25 ms of stance $[m]$	7.4 ± 3.1	14.1 ± 4.9	0.05
Vertical oscillation of COM $[m]$	0.091 ± 0.008	0.116 ± 0.014	0.01
$SL\,[m]$	1.172 ± 0.089	1.325 ± 0.089	0.02
$SR\,[Hz]$	2.94 ± 0.02	2.56 ± 0.2	0.02



(a) (b) (c)

Figure 2.7: Foot strike: a) rear b) mid c) fore; (Lohman *et al.*, 2011). None of the vertical GRF expose an increase during the midstance but a decay suggesting that the runner does not push off the ground.

Table 2.3: SL and SR at different speeds accord. to Elliott & Blanksby (1979)

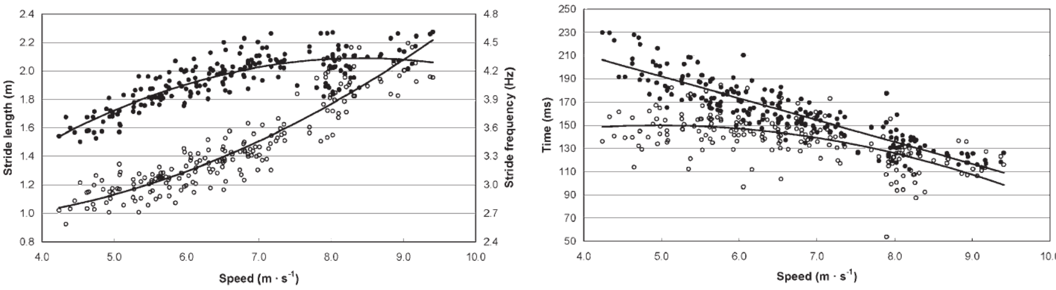
		speed			
	$n = 10$	$2.5\,m\,s^{-1}$	$3.5\,m\,s^{-1}$	$4.5\,m\,s^{-1}$	$5.5\,m\,s^{-1}$
$SL\,[m]$	Males,	0.97 ± 0.03	1.30 ± 0.08	1.56 ± 0.06	1.79 ± 0.10
	Females	0.93 ± 0.07	1.22 ± 0.10	1.44 ± 0.10	1.60 ± 0.08
$SR\,[Hz]$	Males	2.59 ± 0.08	2.71 ± 0.17	2.90 ± 0.12	3.08 ± 0.17
	Females	2.68 ± 0.21	2.88 ± 0.22	3.14 ± 0.10	3.45 ± 0.16

Saziorski *et al.*, 1987), above SR, see fig. 2.8, a. As long as SL does not reaches the plateau (here at approx. $7.5\,m\,s^{-1}$), there might be still enough room for variability of the movement and time to act within this range of motion. In this lower range of speed, more use of slow twitch fibers improve economy. The more the speed increases, the more the economy decreases to toss the legs within a more and more exhausted range of motion, thus SR compensates and the more fast twitched fibers work. The study of Elliott & Blanksby (1979) examined freely chosen SL and SR in skilled but not competitive runners. Tab. 2.3 illustrates the results of SL and SR as a function of speed. Once again, the study of Weyand *et al.* (2000) will be consulted. They examined the ways to increase top speeds which can be theoretically achieved by 1) increasing SR 2) traveling further during flight/increase "force applied to oppose gravity during foot ground contact“ (p.1992) and 3) traveling further during stance/increasing contact length. Eq. 2.11 expresses these three mechanical means. The potential of SR is limited as aforementioned and due to the swing time which is "two-thirds of stride time and therefore the primary determinant of the stride frequency“ (p.1997) i.e., the time for re-positioning the leg is short. CT and FT decrease with an increase in speed, see fig. 2.8, b. Brief contact times are not only associated with an effective stretch-shortening cycle but also with the short time for the re-positioning of the legs.

$$v_f = SR \cdot \frac{F_{avge}}{BW} \cdot l_{stance} \quad v_f \ldots \text{velocity in forward direction} \tag{2.11}$$

$$F_{avge} \ldots \text{averaged vertical GRF}, \quad BW \ldots \text{body weight (force)}$$
$$l_{stance} \ldots \text{distance of CM traveled during stance}, \ \tfrac{F_{avge}}{BW} \ldots \text{average mass-specific force}$$

Fast and slow runners did not show a significant difference in swing time. The mean swing time was $0.332\,s$. Although at top speeds, faster runner exhibited greater stride



(a) (b)

Figure 2.8: SL (dots) vs SR (open circles) and CT (dots) vs FT (open circles) (Nummela *et al.*, 2007)

rates (1.8 Hz to 2.4 Hz), there was a weak correlation between top speeds and maximum stride frequencies. Dillman (1975) reported that ”[ST] decreases as speed of running is incremented” (p.207), because SR, the inverse of ST, increases. At slow speeds, ST is about 0.38 s. The contact lengths were significantly but slightly greater in the faster runners while the flight time (FT) in both groups were 0.128 s. The resultant SL were 1.69 times longer than in the slow counterparts. Although the swifter runners exhibited greater values in the most parameters, the investigators defined the mass-specific force applied to the ground which was 1.26 greater in the swifter runners as the dominant mechanism to improve top speeds. ”Altering the support force applied by only one-tenth of one body weight is sufficient to alter top speed by one full meter per second“ (p.1997). FT and speed have a curvilinear relationship. FT increases at middle to fast speeds, at very fast speeds, FT decreases (Dillman, 1975).

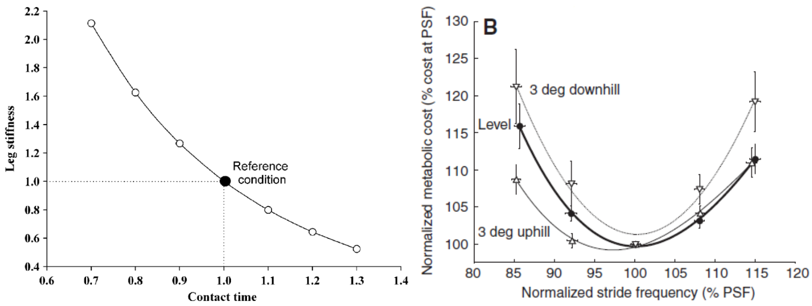
The inverse of the contact time (CT^{-1}) is also referred to ”as the rate of force application“ and is considered as a ”determinant of the energetic cost of running across speed in various species of terrestrial mammals“ (Morin *et al.*, 2007, p.3341). Variations in SR lead to a change in the leg stiffness which is linked to efficiency of the ’bouncing system’. CT is stronger related to the leg stiffness than SR and has a $1 : 2.5$ effect on leg stiffness, for example, ”10 % decrease in CT leading to a 25 % and vice versa“ (p.3342). Changes in CT induced $90 - 96$ % of the variance in leg stiffness, see fig. 2.9, a. Leg stiffness seems to be approx. the same at a wide range of speeds suggesting that there is a preference for a small range of resonant frequencies which are economic. As it had been mentioned, there is an U-shape relationship between SR and the metabolic cost, see fig. 2.9, b. The duty factor (DF) has been mostly used to describe the relation between CT and ST, see eq. 2.12. For example, there is a sudden change in DF while shifting between walking and running (Farley & Ferris, 1998). The above-mentioned study of Morin *et al.* (2007) and Chang & Kram (1999) also made use of the dimensionless factor with showed significant changes during altered conditions of running.

$$DF = \frac{CT}{CT + FT} \qquad DF \ldots \text{Duty Factor} \tag{2.12}$$

To the knowledge of the author, there was only one study examining the effects of both SL and SR "on the generation of stride variability in the human gait" of walking—Danion *et al.* (2003). They described SL and SR as two fundamental (independent) parameters which both have influence on spatial and temporal variability. Any combination other than the preferred resulted in an increase of stride variability. In walking, a SR of about $1\,Hz$ caused lowest variability. Above and under this frequency, there was an increase in variability. In contrast, the "effect of the stride length was rather monotonic" (p.76) and the preferred speed did not result in lowest variability. Larger strides were associated with less variability due to the littler degree of freedom. With this in mind, adjustments are sensitive and further investigations in running may help to unravel this enmeshing.

2.1.3 Anthropometry

The influence of anthropometry on the biomechanics and likewise on economy and performance has been mainly studied through body weight (BW), body height (BH) and leg length (LL). In the following, the usefulness of such anthropometric data for the application in an MC system supporting runners in terms of performance and injury prevention will be examined. First, a non-dimensional parameter will be introduced that allows for comparing of gait pattern inter and intra-individually, even between species. Second, the relations of anthropometric parameters to the predictability of optimal adjustments will be scrutinized.



(a) (b)

Figure 2.9: a) CT vs leg stiffness (Morin *et al.*, 2007) b) preferred SR (Snyder & Farley, 2011)

In the 19th century, the naval architect William Froud found a way to compare "boats of different hull length" (Vaughan & OMalley, 2005, p.350) with respect to dynamic similarity. Different boats caused different wave patterns at the same speed. Further, he recognized that there were speed combinations in which the wave patterns were nearly the same—as "the ratio of the velocity squared to the hull length was the same for both large and small hulls" (p.351). After him, D'Arcy Wentworth Thompson experienced the more general meaning of that scaling law, especially in considering the locomotion of animals "living under the direct action of gravity" (p.353). Finally in 1976, Robert McNeill Alexander "ensured that the Froude number can now take its rightful place as an important parameter for us to employ when studying bipedal gait" (p.354). Eq.2.13 is the modified Froud relation for the application on bipedal movement, whereas eq. 2.14 determines the Froud velocity i.e., Froud number is equal to one. The bipedal gait of walking and running is considered as an inverted pendulum.

$$Fr = \frac{v}{\sqrt{g \cdot LL}} \quad Fr \ldots \text{Froud number,} \quad LL \ldots \text{leg length} \tag{2.13}$$

$$v_{Fr} = \sqrt{LL \cdot g} \quad v_{Fr} \ldots \text{Froud velocity} \tag{2.14}$$

The dynamic similarity expressed by the Froud number predicts that "animals and humans traveling with equal Froude numbers will use the same gait" (Romanov & Fletcher, 2007, p.438). The walking speed is about $0.42\,v_{Fr}$ while running is initiated at $0.7\,v_{Fr}$ (England & Granata, 2007). The maximal walking speed is achieved at $1.0\,v_{Fr}$. Speeds greater than v_{Fr} convey a double flight phase i.e., running gait is unavoidable. Geometrically similar movements cause similar dynamics. Alexander established the following empirical relationship (p.354):

$$\frac{SL}{LL} = 2.3\frac{v^2}{(g \cdot LL)^{0.3}} \tag{2.15}$$

This means for running that two runners with different LL but the same ratio of SL/LL are also dynamical similar (Romanov & Fletcher, 2007). Vaughan & OMalley (2005) did not specify the range of validity for eq. 2.15, but should be in the range where SL is dominant i.e., walking to middle fast speeds. Romanov and colleges stated that the Froud-based approach ignores the role of muscles and tendons that mainly act vertically and, therefore, proposed also to consider the vertical Froud number. This more detailed approach leads to the Strouhal number. However, to the knowledge of the author only the Froud number has been used in the analysis of walking and running gait.

The Froud number allows for comparison of dynamic similarity and hints that an increase in speed is associated with an increase in SL, and further, that LL and SL are positive related i.e., longer strides lead to greater SL. Nonetheless, the Froud approach has a more general meaning and thus has been applied comparing the gait of various animals. A closer review revealed that other anthropometric parameters than LL are suggested to have an influence on SL and SR, economy, and performance, too. In sprint running, Dillman (1975) reported a strong correlation of the body height (BH) and LL to SL. Cavanagh & Williams (1982) confirmed that there is a strong correlation between LL and SL at least for fast running speeds. They expressed the optimal SL (minimal metabolic cost) as a multiple of LL, see eq. 2.16. The range of this multiple was between 1.30 and 1.65 i.e., there was a "considerable variability among subjects" (p.33). In contrast to these findings, Svedenhag & Sjodin (1994) stated a poor relation of step length to economy.

$$SL_{opt} = 1.40 \cdot LL \tag{2.16}$$

Citing Scholich (1978) Buckalew *et al.* (1985) reported that a "mean of 1.15 times the standing height as being the approximate value for male distance runners", whereas Cavanagh *et al.* (1985) found SL of two elite runners ranging from 100 % to 115 % of their body height. Of note is that in the study of Cavanagh & Williams (1982) seven of ten fit subjects (running 64 to 177 km per week, $\dot{V}O_{2max} = 64.7\,ml\,kg^{-1}\,min^{-1}$) were over-striding, while the remainder were using shorter than optimal strides. Cavanagh & Kram (1989) found low correlations ($r = 0.36$) between the anthropometric variables such as BH, LL, the limb segment mass, and the stride variables such as SL and SR. Svedenhag & Sjodin (1994) compared elite runners of middle and long distances in their relations of step length to anthropometric measures. Step lengths at speed $15\,km\,h^{-1}$ and $18\,km\,h^{-1}$ "did not differ significantly between the groups" (p.305). The middle-distance runners initiated a raise in speed dominantly with an increase in SL, whereas the long-distance runners in SR. In this study, step length was positively related to BH, and negatively to LL. Bangsbo & Sjogaard (2001) findings were similar in elite runners. They postulated a positive relationship of step length to BH and a negative relationship to LL/BH quotient. The last finding is obviously conflicting with the Froud approach. The parameter LL is still of debate concerning an optimal SL. The intuitive assumption that a longer LL "would be best suited by longer strides" (Cavanagh & Williams, 1982, p.34) cannot be hold over all speeds. Sprinters are usually short-legged, as are long-distance runners, whereas middle-distance are long-legged. At high speeds as required in sprinting, smaller LL might be of advantage due to the relative muscular body composition of sprinters (Ross & Ward, 1982; Martin & Morgan, 1992) i.e., sprinters are heavier and have a greater Body Mass Index (*BMI*, *BW* divided by the second power of BH) than distance runners. Shorter LL facilitate smaller moments of inertia while tossing the leg. In contrast, Marathon runners have a slighter physique and concentrate segment mass closer to the joints, or closer to the hip. The review on running and science of Bangsbo & Sjogaard (2001) showed that women distance runners are relatively long-legged, while Anderson (1996) stated that "women distant runners are taller and lighter than average“ (p.79). Nonetheless, according to Elliott & Blanksby (1979), "longer anthropometric segements resulted in decreased SR" (p.15) and there was a close inverse relationship between height, LL and SR. At speeds above $7.5\,m\,s^{-1}$, the influence of LL decreased, but in shorter runners "continued to increase SL“ (p.15). Tab. 2.4 holds the multipliers of LL and BH for the calculation of SL in sprint running. There the original references were given cited in this study through Hay (1973). Tab. 2.5 shows the ratio of SL to BH, while tab. 2.6 illustrates the ratio of SL to LL. Both ratios were given for four speeds of recreational

Table 2.4: SL in sprinting as a function of LL and BH

Study	Gender	Multiplier of LL	Multiplier of BH
Hoffmann (1965, 67)	Female	2.16	1.15
Hoffmann (1965, 67)	Male	2.11	1.15
Rompotti (1967)	n.a.	n.a.	1.17 ± 0.102

Table 2.5: Ratio SL to BH accord. to Elliott & Blanksby (1979)

	Speed			
	$2.5\,m\,s^{-1}$	$3.5\,m\,s^{-1}$	$4.5\,m\,s^{-1}$	$5.5\,m\,s^{-1}$
Males, $n = 10$	0.54 ± 0.03	0.72 ± 0.06	0.87 ± 0.03	1.00 ± 0.05
Females, $n = 10$	0.55 ± 0.05	0.75 ± 0.06	0.86 ± 0.05	0.96 ± 0.05

runners on a treadmill. Cavanagh *et al.* (1977) compared good and elite runners at $4.97\,m\,s^{-1}$ and found that SL and LL were correlated with $r = 0.67$ in the elite runners and with $r = -0.10$ in the good runners. The good runners preferred longer strides, whereas the elite runners performed SL better related to their LL. Elliott & Blanksby (1979) alluded that the weak to moderate correlations might be improved when the best runners are isolated. For example, a correlation of SL to BH found by Hoffman (1965) cited in Elliott & Blanksby (1979) was 0.69. After having isolated the best fifteen runners, the correlation raised to 0.82.

Landers *et al.* (2011) studied runners (age $27.2 \pm 3.0\,yrs$) in a world champion triathlon event. They found a strong relationship between SL and BH (speeds approximately $5\,m\,s^{-1}$). Taller runners used longer SL and slower SR. Faster runner also used longer SL than slower runner. SR had no significant correlation to the running speed. Landers and colleges supposed that heavier runners would use lower SR but had to reject this hypothesis. They reasoned that a similar body shape and the similar velocity of contraction have stronger influence than BW. Dillman (1975) found

Table 2.6: Ratio SL to LL (trochanterion) accord. to Elliott & Blanksby (1979)

	speed			
	$2.5\,m\,s^{-1}$	$3.5\,m\,s^{-1}$	$4.5\,m\,s^{-1}$	$5.5\,m\,s^{-1}$
Males, $n = 10$	1.10 ± 0.09	1.46 ± 0.09	1.71 ± 0.07	1.96 ± 0.08
Females, $n = 10$	1.11 ± 0.10	1.48 ± 0.13	1.70 ± 0.14	1.89 ± 0.16

a low inverse relationship between SL and BW that might be due to higher work to lift CM. Ross & Ward (1982) reported a low inverse relationship of lower proportional BW to the increasing distance of running event. The addition of weights up to 1.1 kg at each ankle did not change SL and SR behavior (Cavanagh & Kram, 1989). The more a weight is added towards CM, the less its influence is on the stride parameters (Anderson, 1996). Nonetheless, BW has been used to normalize, especially the aerobic demand. The underlying assumption is that the aerobic demand is independent of BW. If there are large variations of BW in a group of runners, then this normalization is not justified, because larger individuals have a lower-mass specific cost of locomotion. The amount of the valid variability could not be unveiled. Saunders *et al.* (2004) could prove that BW and maximal thigh circumference ($r = -0.58$) are related to economy. Regarding Saunders *et al.* (2004), heavier athletes have less metabolic cost per kg of body mass. They emphasized that runners concentrating less mass in the lower limb segments waste less energy moving these segments. Cavanagh & Kram (1989) investigated high relations of LL, pelvic width, foot length to economy. The findings of Anderson (1996) are in consensus with the previously mentioned study. Foot length was found negatively correlated with running economy in elite runners. They also found a low correlation between BH and performance, and between Marathon time and the *BMI*. Higher ponderal indexes (BW divided by BH) were observed in elite and good runners. In elite runners, BH and BW were smaller in comparison to the good and average runners. Hips and shoulder are suggested to have no relationship or a moderate negative correlation to economy. Ross & Ward (1982) experienced narrower hips in 400 m runners than in the distance runners but "no consistent pattern of significant differences among the events [...] in the proportional length, breadths and skin fold thickness“ (p.128).

In triathlon running, anthropometry accounts for 47 % of the variance in performance according to Saunders *et al.* (2004). It has been shown that anthropometry has an influence on SL and SR, and economy and performance. Anthropometry offers a range of possible adjustments of SL and SR to a speed, while there are mechanical efficient and physiological optimal ones that both have impact on performance. Runners, especially recreational runners, do not fully use their anthropometric potential. Poor SL, especially the over-striding, is not just inefficient in means of metabolic or mechanical

Table 2.7: Optimal SL accord. to Cavanagh & Williams (1982)

Subject	Mass $[kg]$	Height $[m]$	Leg length $[m]$	$\dot{V}O_{2max}$ $[ml\,kg^{-1}\,min^{-1}]$	Chosen $\dot{V}O_2$	Chosen SL $[m]$	Optimal $\dot{V}O_2$	Optimal SL $[m]$	$\triangle\,SL$ $[m]$
1	72.6	1.841	0.953	59.4	41.2	0.1281	41.2	1.252	0.029
2	73.9	1.864	0.969	62.3	45.4	0.1225	45.2	1.262	0.037
3	68.5	1.791	0.938	68.2	45.9	0.1412	45.8	1.375	0.037
4	73.9	1.791	0.934	65.0	45.8	0.1352	45.3	1.261	0.091
5	66.7	1.803	0.949	64.8	48.2	0.1290	48.0	1.327	0.037
6	70.8	1.810	0.931	73.3	52.9	0.1373	52.7	1.316	0.057
7	66.2	1.753	0.894	64.7	45.2	0.1297	44.4	1.231	0.066
8	80.3	1.829	0.953	72.3	44.9	0.1296	44.8	1.272	0.024
9	70.8	1.753	0.910	54.7	43.3	0.1281	43.2	1.308	0.027
10	76.7	1.715	0.842	62.0	46.6	0.1405	46.6	1.391	0.014
Mean	72.0	1.795	0.927	64.7	45.9	1.321	45.7	1.299	0.042
SD	4.4	0.045	0.037	5.6	3.1	0.061	3.1	0.054	0.023

meanings, but can also cause high joint loading and lead to a higher risk of injury, especially under fatiguing conditions. Nonetheless, anthropometric parameters are not suggested to "be used to accurately predict or prescribe SR or SL on an individual basis“ (Cavanagh & Kram, 1989) but might serve as reference values.

2.1.4 Shoes and environment

In this subsection, the influence of shoes, cushions and the environment on economy, performance and the risk of injury will be discussed. This subsection accentuates those biomechanical factors which connect the runner with the environment. These factors have also been found to restrict the variability in economy, performance and the risk of injury.

Running shoes are available in many fashions, see fig. 2.10. This is not only due to the fact that there are several use-cases such as disciplines in running, indoor and outdoor conditions, but also the different anatomical conditions of runners. Walther (2004) reports that the "concepts used in sport shoe design are currently in a phase of a fundamental change" (p.167). Same intentions (i.e., improving performance or lowering risk of injuries) resulted in different shoe designs. The controversy may be

Table 2.8: Biomechanical factors related to better economy in runners accord. to Morgan *et al.* (1989, p.86)

Average or slightly smaller than average height for men and slightly greater than average for height for women	Leg morphology which distributes mass closer to the hip joint
Low percentage body fat	Narrow pelvis
High ponderal index and ectomorphic or mesomorphic physique	Smaller than average feet and lightweight but well-cushioned shoes
Stride length which is freely chosen over a considerable training time	Low vertical oscillation of body center of mass
More acute knee angles during swing	Less range of motion but greater angular velocity of plantar flexion during toe off
Arm motion which is not excessive	Low peak ground reaction forces
Faster rotation of shoulders in the transverse plane	Greater angular velocity excursion of the hips and shoulders about the polar axis in the transverse plane
Effective exploitation of stored elastic energy	More comprehensive training history
Running surface of intermediate compliance	

(a) (b) (c)

Figure 2.10: Shoe types: a) rocker bottom, b) standard, c) minimalistic; (Sobhani *et al.*, 2013)

pictured with two extremes. On the one hand side, there is footwear such as the 'rocker bottom shoes' which are very stable and allow for a high cushioning and on the other hand side, there is the barefoot running, which requires stabilization and cushioning through the muscles. It has been shown that assumptions in both types could not be fully proven true.

The first wave of shoe-wear development, especially in running, prioritized the management to "shock absorption and control/stabilization" (Novacheck, 1998, p.91). It was believed that high impact forces and the deviation from an anatomical correct alignment of the skeleton cause injuries. Further, it has been assumed that the properties of shoes and surfaces have an effect on the reduction of those causes. This concept holds tight to three principles, which were 1) supporting 2) cushioning and 3) guidance (Walther, 2004). In this theory of running, principle one aimed at accomplishing the 'natural ride', whereas principle three attempted to guide the lower extremities within an allowed deviation from the skeletal alignment. For example, the excessive pronation has been suspected to compromise the ankle joint. Cushioning (principle two) was seen to be most important to reduce the peak impact loads.

Rearfoot striking is prevalent in all runners. Novacheck (1998) quantified rearfoot runners at 80 %. Bertelsen *et al.* (2012) cited several studies that showed different proportion of rearfoot, midfoot and forefoot runners at different levels. For example, a study with elite and sub-elite runners revealed 74.9 %, 23.7 % and 1.4 %, respectively, and a study with recreational and sub-elite runners revealed 94.4 %, 3.6 % and 1.9 %, respectively. Based on these findings the portion of the midfoot strikers tends to increase in the elite runner. In the study of Bertelsen *et al.* (2012) 98.1 % of inactive persons taking up running were rearfoot strikers. A total of 903 subjects took part, whereas 456 subjects were male and 447 subjects were female. Males had a proportion of 96.9 %, 0.4 % and 0.9 %, respectively, whereas 1.8 % had a mixed pattern. Among women the proportion was 99.3 %, 0 % and 0 %, whereas 0.7 % had a mixed pattern. If shoes really can constrain the foot, then all these numbers might represent the striking patterns of standard running shoes of the first wave. Nonetheless, due to the dominance of the heel-striking Winter & Bishop (1992) cited in Novacheck (1998, p.92) summarized the supporting-cushioning-guidance concept in a chronological order:

1. Shock absorption at heel contact reducing the initial spike of reaction force (protects against joint cartilage damage).
2. Stance phase—protects against the rough ground surface.
3. Aligning the forefoot to achieve a uniform force distribution at the major chronic injury sites.

Both shock absorption and stability cannot be maximized in one shoe or in one of its regions, because these principles are oppositional. In order to enforce stability, the shoe has to be designed stiffer, whereas cushioning requires thicker-soled or softer materials. The shock attenuation between different running shoes has been shown to differ in about one-third (Novacheck, 1998). Further, after approximately 80 km the attenuation is reduced about 25 % and levels out at about 33 % reduction after 160 to 250 km. Liang & Chiu (2010) made similar disclosures while measuring the reduced cushioning abilities to up to 300 km. Running shoe manufacturers endeavor to establish the properties in the cushioning system which allows for a recoil effect as it is known from the soft tissue. This amount of returned energy is quite small compared to the stretch-shortening cycle. However, the peak impact loads at the heel strike occur within the first 50 ms. They have been associated with injuries at the skeletal system through models of animals and theoretical reasoning (Walther, 2004).

The cushioning system of the shoe may have little influence on the resulting loads in the body (Taunton *et al.*, 1988; Lane & Bloch, 1986). The loads are suggested to be

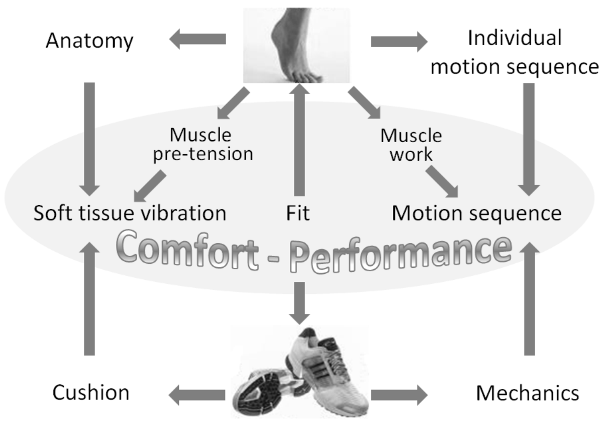


Figure 2.11: Interaction between runner and shoe (adapted from Walther (2004))

dependent from the speed of the runner while this influence can cause a range of 100 % of the variability in the loads at impact. Runners with high loads i.e., high vertical *GRF* rarely suffered more injuries than runners with low loads. Runners have no higher incidence for injuries than non-runners. Running on rough surface does not cause more injuries than running on softer surfaces. Low, middle and high loads at impact are not of importance for the risk of injury. Therefore, it has been reasoned that impact loads do not serve as a predictor for injuries in running. On the contrary, around the push-off event three to five times higher loads than at the initial contact have been detected, see also fig. 2.7. Furthermore, the muscular-skeletal system seems to be prepared for high loads within a genetically determined range. The tissue adapts to the load within this range. For example, high impacts stimulate the growth in bone density. If the time for recreation is too little, then stress fractures in bones can occur. Indeed, overuse injuries usually occur in the forefoot (Sobhani *et al.*, 2013). These new insights contradict both designs of the standard cushioning and the minimalistic shoe. The intention to reduce the high impact loads due to the risk of injury might not be justified, neither with shoe cushioning nor with shoes supporting the forefoot striking. In 1991, Robbins & Gouw alluded that "modern athletic footwear is unsafe because it attenuates plantar sensation that induce the behavior required to prevent injury" (p.218). Even years later, Novacheck (1998) concludes that "[despite] the developments in shoewear technology, the overall rate of injury in distance runners has not changed significantly" (p.92).

The traditional belief was that shoes and surfaces determine the movement of the lower limbs. Therefore, developments were engaged to enforce an optimal movement of the lower limbs. Novacheck (1998) emphasizes the role of neurologic control—its influence on lower limb movements. Walther (2004) argues that a runner can apparently adjust one's running technique to the shoe and surface in a way that keeps the runner as near as possible to the individual movement pattern. This happens through adjustments in the lower limb geometry and the leg stiffness. Each joint as well as all conjunctions between rigid and soft tissue contribute to this adjustment. For example, mechanoreceptors measure pressure distribution and the kinematic of the foot and send afferent information to higher control structures. The muscles act as the control and motor system, in order to evoke a proper stiffness in the lower limbs through tension. The pre-tension has a crucial function because it prepares the stiffness before the impact happens. Impacts cause vibrations in the soft tissue. These vibrations can be perceived as discomforting and even painful. In order to cushion these vibrations, the muscles counteract with tension. The higher the tension, the higher the energy expenditure in running. The energy expenditure was shown to be different in several types of shoes and surfaces. In general, soft and viscous materials led to less work than the harder and elastic ones. For example, in treadmill running the oxygen consumption could be reduced by 2.8 % by wearing well-cushioned shoes rather than stiffer shoes of equal weight (Anderson, 1996). Cushioning can increase the weight of the shoes that also would cause higher running cost. With this in mind, higher impact forces may induce high soft tissue vibrations, which effect higher work rates, and finally, influence comfort, fatigue and performance. Therefore, the second wave of shoewear development focuses on low soft tissue vibrations with the associated low muscle pre-tension, see fig. 2.11.

The traditional concept attempted further at the optimal alignment of the lower limbs e.g., with the help of inserts. Injuries were associated with the static and dynamic alignment of the skeletal system. For example, an excessive heel varus or valgus was considered to be responsible for overload damages in the ankle joint (for details see sec. 2.1.6). Nevertheless, the movement pattern of a runner cannot be significantly changed through shoes, inserts or the surface. The movement pattern might be characterized by a high constancy. There have been attempts to explain this. Wilson *et al.*

(1996) cited in Walther (2004) reasoned that the movement pattern tends to the minimal required mechanical work. In this line, Nigg (2001) assumed that the conditioned neuromuscular system tries to avoid deviations from the individual pattern. If a shoe supports the individual movement of a runner, then a minimal work rate is expected. The parameters fatigue, comfort, muscle work and performance should then express this relation, see fig. 2.11. Wallack (2004) reports that there were studies that could confirm this relation. For example, movements in comfortable and thus well-fitting shoes and barefoot caused similar patterns and oxygen expenditures. Modern developments in shoewear regard individuality in movement by trying to group individual patterns. However, it remains elusive to prescribe running shoes even when regarding individuality (Richards *et al.*, 2009). Moreover, even if unshod running would be an advance delivering higher performance and lower risk of injury, the neuromuscular system is conditioned with shoes in the majority of runners. The influence of shoes and surface cannot be neglected but an individual solution seems to be necessary in which the comfort, hence the feeling, may play a distinctive role. This new paradigm is listed according to (Nigg, 2001, p.7):

1. Forces acting on the foot during the stance phase act as an input signal.
2. The locomotor system reacts to these forces by adapting the muscle activity.
3. The cost function used in this adaptation process is to maintain a preferred joint movement path for a given movement task (e.g., running).
4. If an intervention supports the preferred movement path, muscle activity can be reduced. If an intervention counteracts the preferred movement path, muscle activity must be increased. An optimal shoe, insert, or orthotic reduces muscle activity.
5. Thus shoes, inserts, and orthotics affect general muscle activity and, therefore, fatigue, comfort, work, and performance.

2.1.5 Physiology

Although the purpose of the thesis is to establish a biomechanically grounded MC system, it has been found of importance to consider physiological aspects, too, in order to better estimate the advantages and drawbacks of the project. Some aspects of the physiological system such as the metabolic cost and the maximum oxygen consumption have been mentioned in the previous sections. For example, metabolic cost has been mostly used to describe economy in running, and heart rate based approaches help to estimate the state of the runner and to optimize training sessions. The key question of this section addresses how do physiological factors determine running and how useful are physiological parameters for prediction and observation of endurance performance or fatigue, respectively. It is not the intention to give a full review on physiology and its relation to performance but rather to highlight important physiological factors and parameters, which may support to profile runners, and allow for evaluating the conditions under which potential biomechanical adjustments occur.

This review follows Burnley & Jones (2007), Karp (2008) and Bangsbo & Sjogaard (2001) and also starts with the traditional view on the determinants in pyhsiologically based performance diagnostic which are 1) the maximal oxygen uptake ($\dot{V}O_{2max}$), 2) the lactate threshold (LT), 3) the running economy, and less mentioned 4) the critical power (power output at the maximal lactate steady state), see fig. 2.12. The maximal oxygen uptake has been seen as a good indicator of endurance in performance in heterogeneous groups, whereas in good and especially in elite runners this parameter being high "becomes a less good predictor" for the endurance performance (Bangsbo & Sjogaard, 2001, p.22). Among elite runners with similar performance, $\dot{V}O_{2max}$ "may vary dramatically“ (Bangsbo & Sjogaard, 2001, p.22). The highest value in male runners is that of 10 km, world record holder Dave Badford with 85 $ml\,kg^{-1}\,min^{-1}$ and in female runners is that of Joan Benoit, winner of the inaugural 1984 Women Olympic Marathon, with 78 $ml\,kg^{-1}\,min^{-1}$ (p.21). There are very successful athletes with lower values e.g, Derek Clayton, world marathon record holder in 1969, with a value of 69 $ml\,kg^{-1}\,min^{-1}$. Values in healthy young men are between 45 and 55 $ml\,kg^{-1}\,min^{-1}$. Intensive training can raise $\dot{V}O_{2max}$ by 20 to 25 % but in the average runners cannot go up to the values of elite runners due to hereditary factors. In contrast, in the elite runners $\dot{V}O_{2max}$ did not increase over the years. A higher than average maximal oxygen uptake seems to be necessary but more important to performance are other factors. For example, two elite runners may have the same $\dot{V}O_{2max}$. The runner with the lower oxygen uptake at a given speed is more economical and, therefore, might reach higher

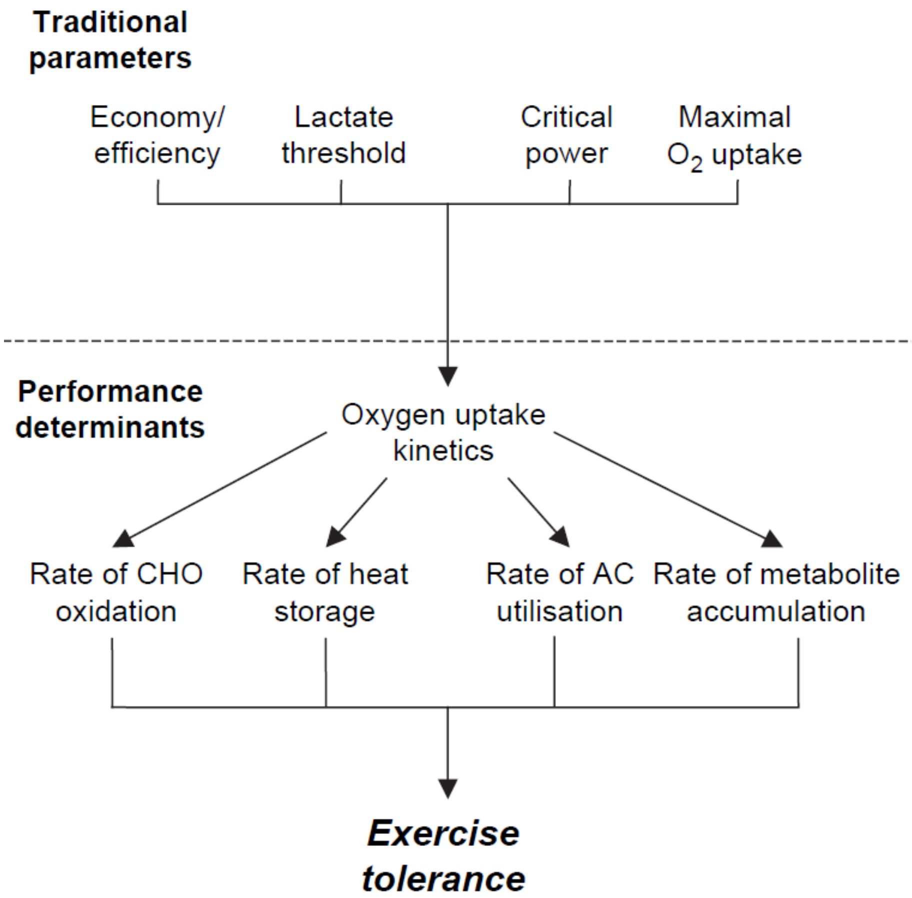


Figure 2.12: Physiological determinants of endurance performance (Burnley & Jones, 2007). The traditional parameters place constraints upon the kinetics of VO_2 but cannot be purely used for prediction performance among elite runners. Outstanding athletes only might be recognized by those parameters. Otherwise, it is more the efficiency of the kinetics of VO_2 and its specialization to a certain distance, and thus to a certain exercise tolerance/fatigue resistance. Improvements in performance are not inevitably related to changes in the traditional parameters. The physiological perspective considers some determinants for the oxygen uptake kinetics. The rate of carbohydrate oxydation (CHO), the rate of heat storage, the rate of anaerobic capacity (AC) utilization, and the rate of of metabolic accumulation have been associated with an improved performance.

peak speeds. Another example is given in fig. 2.13. Runner A has highest $\dot{V}O_{2max}$ and reaches as runner B having a smaller $\dot{V}O_{2max}$ the same peak speed. Runner B is more economical than A. Despite runner C has the lowest $\dot{V}O_{2max}$, this runner is the most economical and, therefore, is the fastest. Indeed, it is well-documented that the running economy and performance are strongly correlated (Saunders *et al.*, 2004). The best runners are usually the most economical.

In the last example, the longer the distance would be, the more it could be ensured that the most economical i.e., runner C, would win the race. Running economy is referred to the oxygen uptake at sub-maximal running speeds i.e., when the aerobic system is still dominant, see fig. 2.15, a. This is justified, because most of the time in middle- and long-distance race running is at submaximal speeds. Nonetheless, different combinations are possible—runners with a better economy and a low $\dot{V}O_{2max}$, and vice versa. It has also to be mentioned that there are studies, which did not find such a strong correlation between the running economy and the performance e.g, Williams & Cavanagh (1987) with heterogeneous runners. Saunders *et al.* (2004) argued that the prediction of the performance in races at fast speeds through the economy improves while observing homogeneous runners, especially at slower speeds. $\dot{V}O_{2max}$ may be helpful to estimate the potential of runners and to assort them but not in predicting the fastest runner in middle- and long-distance running of a homogeneous group. "An outstanding runner may have good or excellent values in both running economy and $\dot{V}O_{2max}$" (Bangsbo & Sjogaard, 2001, p.87). As a practical solution, it has been suggested to use the peak running speed collected e.g., in a ramp test, as a predictor of performance (p.26). Another way is to calculate the individual aerobic running capacity, which is the fractional utilization of the $\dot{V}O_{2max}$ at selected running speeds. The $\%\,\dot{V}O_{2max}$ value has shown a good correlation to performance at sub-maximal speeds, especially in heterogeneous groups, and therefore, it has been used as a reference value to prescribe training intensities—not without critics, as discussed later in this chapter.

In order to better compare individuals, oxygen uptake has often been normalized for the body weight (BW). Oxygen uptake and BM are not proportionally related and differ between animals; and "in humans the oxygen uptake per kilogram of BM is higher in children than in adults" (Saunders *et al.*, 2004, p.469). The inverse relationship has

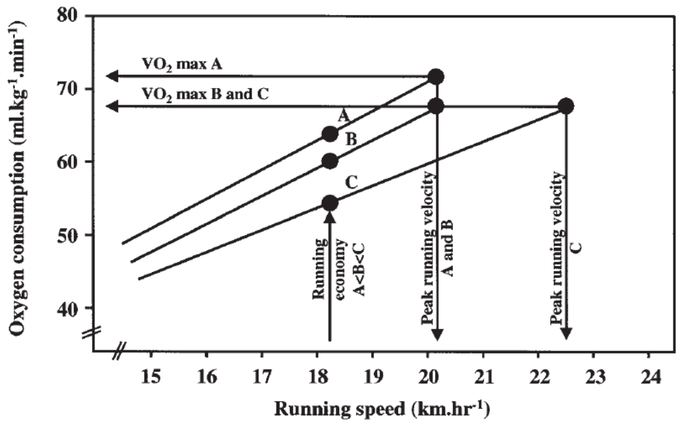


Figure 2.13: $\dot{V}O_{2max}$ and running economy (Bangsbo & Sjogaard, 2001)

been numbered from 0.66 to 0.76. According to Bangsbo & Sjogaard (2001), the ratio of 0.75 will be taken for further considerations i.e., oxygen uptake is normalized with 0.75 of BW ($kg^{-0.75}$). Tab. 2.9 compares two runners of different body masses, $\dot{V}O_2$ at speed $18\,km\,h^{-1}$ ($\dot{V}O_2$-18) and $\dot{V}O_{2max}$. The lighter Runner B has a higher $\dot{V}O_2$-18 with kg^{-1} and, therefore, B seems to be less economical than A. The runner B would be referred to a training improving one's economy. Taking $\dot{V}O_2$-18 with $kg^{-0.75}$ into consideration reveals that both runners do have a similar economy. For $\dot{V}O_{2max}$, it is the other way round. Here, runner A would be referred to a training that focuses on improving $\dot{V}O_{2max}$. The normalization with $kg^{-0.75}$ gains similar results.

In the traditional concept $\dot{V}O_{2max}$ is an important indicator for an upper boundary for the maximal oxygen uptake. In order to describe training intensities for aerobic and anaerobic demand, further boundaries were needed. LT has been widely used for an upper boundary for moderate (aerobic) exercises, because the lactate threshold is "determined mainly by the oxidative capacity of the skeletal muscle" (Joyner & Coyle, 2008, p.38). The upturn in the lactic acid concentration is about 60 % of $\dot{V}O_{2max}$ in untrained subjects, in contrast to trained subjects can also range between 75 % and 90 % of $\dot{V}O_{2max}$. With an increase in the exercise intensity above LT, lactic acid concentration rises but can be removed to hold a steady state between production and removal until the critical power or maximal sustainable power output has been reached (Burnley & Jones, 2007). Above critical power, the steady state cannot be hold. A

Table 2.9: Running economy in comparison to different body mass normalization (Bangsbo & Sjogaard, 2001)

Runner	Mass	$\frac{\dot{V}O_2-18}{\dot{V}0_{2max}}$	$\dot{V}O_2$-18		$\dot{V}0_{2max}$	
	$[kg]$	$[\%]$	$[\frac{ml}{kg\,min}]$	$[\frac{ml}{kg^{0.75}\,min}]$	$[\frac{ml}{kg\,min}]$	$[\frac{ml}{kg^{0.75}\,min}]$
A	80	75	55.5	166	74	221
B	50	75	61.5	164	82	218

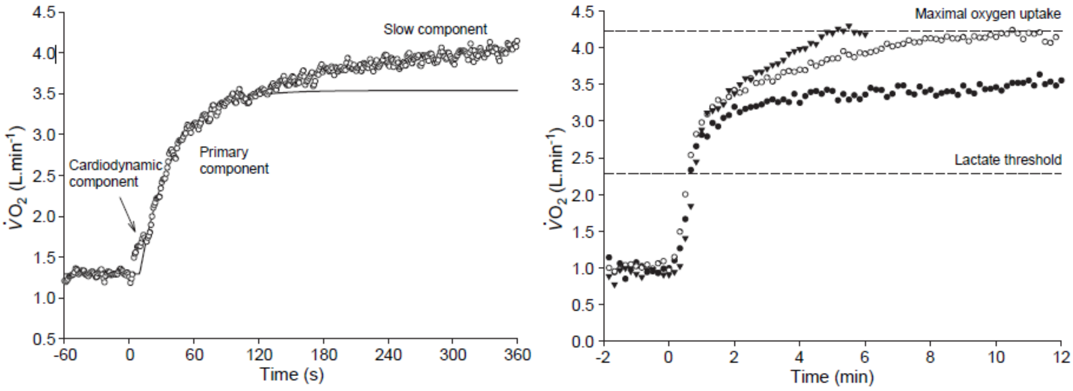
high rate of "ATP[1] is converted to ADP[1] and P[1] to power the cross bridges" in the muscles (Bassett & Howley, 2000, p.80). This drives the anaerobic pathways i.e., results in a "greater rate of carbohydrate turnover, an accumulation of pyruvate and NADH[2] in the cytoplasm of the muscle fiber, and an increase in lactate producation" (p.80). Nevertheless, it is important to mention that running economy, $\dot{V}O_{2max}$, LT and critical power are quite successful in predicting performance, especially in heterogeneous groups, but less in homogeneous groups of good to elite runners. Without going to deep into exercise physiology, some aspects of $\dot{V}O_2$ kinetics are elucidated to show the strength and weakness of such physiological parameters.

The concerns with the traditional view have arrived from the knowledge that these parameters are not suitable to prescribe training intensities for an individual based on fraction of a single parameter (Magness, 2009). Nevertheless, the kinetics of $\dot{V}O_2$ is determined by these parameters, which "place constraints upon" (Burnley & Jones, 2007, p.63) it, but it is the economy of the oxygen utilization that plays a crucial role on performance (Karp, 2008). There are three phases of $\dot{V}O_2$ response to exercises being of low to moderate intensity, see fig. 2.14, a. The first one, named the cardiodynamic phase, is characterized by a sudden increase in $\dot{V}O_2$, "chiefly, as a result of increased venous return via the muscle pump on one hand, and increased right ventricular output elevating pulmonary blood flow on the other" (Burnley & Jones, 2007, p.66). Therefore, this increase is not subjected to muscle oxygen uptake that has a delay of about 10 s to 20 s. The next phases are of more importance to the thesis. In phase two, also called the primary component or fundamental phase, an increase of oxygen uptake can be seen in an exponential fashion. This may describe the sub-maximal intensity. Beyond this,

[1]ATP... Adenosinetriphosphat, ADP... Adenosindiphosphat, P... Phosphor

[2]While nicotinamide adenine dinucleotide (NAD^+) becomes reduced during redox reaction, NADH is formed and is further used as a reducing agent.

there can be a slow component of an increase in oxygen uptake. As the intensity rises further e.g., in an incremental exercise test, the athlete approaches to one's maximum level of oxygen uptake ($\dot{V}O_{2max}$). In the last phase, the anaerobic system dominates. For example, the fast twitched muscle fiber types have been associated with the slow component. On one hand side, the anaerobic system ensures a high performance, but the on the other hand this system produces more (fatigue-related) metabolites that cause full-exhaustion i.e., an exercise cannot be held for a long time. This may also be seen as self-protection. One of the most critical circumstance during intensive exercise can be regarded to heat production. Those circumstance have an impact on the exercise tolerance, in other words, fatigue resistance. The traditional parameters can be used to distinguish the work load or the training intensity. The upper boundary of a moderate exercise is LT. There are two components, the cardiac and the primary. The endurance time can be greater than four hours. With time, hyperthermia and a reduced central drive turn up. In a prolonged run, muscle damage can occur and the motivation is handicapped by "central fatigue“. The next intensity category is described by heavy exercises. In this, the boundaries are set at LT and the critical power. This intensity is related to three components or phased of $\dot{V}O_2$. The intensity is incremented gradually in a way that the steady state of the maximal lactic concentration (around critical power) is never broken through. The oxygen uptake is elevated but does not reach its maximum. The work can be sustained for three to four hours as high rates of glycogen depletion and hyperthermia impacts exercise tolerance. If the intensity is increased so that the lower boundary is now set to the critical power, then the exercise can be named severe. There can be two or three components. Usually, a slow component is evident after the primary phase. There is no steady state and $\dot{V}O_{2max}$ as the upper boundary will be reached before fatigue, see the white circles and black triangles in fig. 2.14, b. In severe exercises, the anaerobic system dominates clearly, so that there is strict depletion of finite energy store and an accumulation of fatiguing metabolites. The endurance time can be up to $45\,min$. The highest attainable intensity is here described as the extreme and is usually achieved in sprints. There is no slow component. The time is too short, in which the aerobic system could provide ATP for the cross-bridge mechanism. The anaerobic system is dominant and is responsible for performance. As a consequence, $\dot{V}O_{2max}$ cannot be achieved, see the black circles in fig. 2.14, b. The fatiguing mechanism is the same as in the severe exercise, but additionally, there is

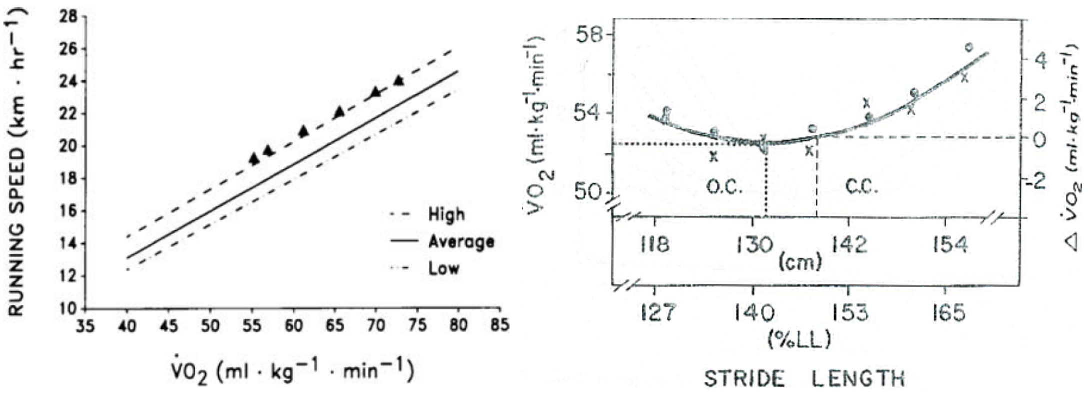


(a) (b)

Figure 2.14: a) The three components of the $\dot{V}O_2$ b) The $\dot{V}O_{2max}$ cannot be reached in short and very intensive exercises. (Burnley & Jones, 2007)

an excitation and contraction coupling failure i.e., the neuromuscluar load is highest. While trying to sort the $5\,km$ run into one of the domains, it is obviously somewhere between severe and extreme one, because the time ranges from about $13\,min$ in elite runners to 30 min in recreational runners. Those runners are successful with middle-distances, which can integrative contribute from the aerobic and anaerobic capabilities (Brandon, 1995).

The analysis of lactic concentration in blood during exercise had been considered a very efficient tool in performance diagnostic. Later on, physiologists have moved away from the former optimism or even the overestimation and criticized this concepts using without further constraints, especially, the daily constitution, beverage, and nutrition (Smekal *et al.*, 2011). The aforementioned concept of exercise intensity can be built up even without the lactate threshold. Therefore, usually a spiroergometry has been used to determine two ventilatory thresholds in a graded and incremental all out test on a treadmill. The first ventilatory threshold (VT_1) describes the point at which "the minute ventilation rate ($\dot{V}_E$) increases disproportionately" (Schwellnus, 2009). This point in time is usually associated with the lactate threshold i.e., when lactic concentration rises exponentially up. To this time point, the relation of the minute ventilation to carbon dioxide ($\dot{V}_E/\dot{V}_{CO2}$) does not change. There is an increase in the carbon diox-



(a) (b)

Figure 2.15: a) VO_2 vs speed (Joyner & Coyle, 2008). Regression lines for high, average, and low running economy in elite endurance runners with similar VO_{2max} and LT. b) VO_2 vs SL (Cavanagh & Williams, 1982). Preferred SL, at approximately $1.4 \cdot LL$, is most economical. Below and above the preferred SL the oxygen uptake increases in an U-shaped fashion.

ide production. When the H^+ ions get released from the lactate, they are banned in a weak acid carbonic conjunction (H_2CO_3). This carbonic acid dissociates into water (H_20) and carbon dioxide ($C0_2$). This increase in the CO_2 stimulates further the rate of breathing, which supports the removal of excess CO_2 levels. Finally, the increase in the minute respiratory rate is accompanied by an increase in carbon dioxide. This proportion changes at the second threshold VT_2. For example, the bicarbonates cannot neutralize the hydrogen ions (H^+) i.e., there is a significant increase in the ratio $\dot{V_E}/\dot{V_{CO2}}$ and likewise with $\dot{V_E}/\dot{V_{O2}}$. Therefore, this change has also been labeled 'respiratory compensation point'. Another association with this time point is the maximal lactic steady state.

Through training, the kinetics of $\dot{VO_2}$ can be improved even without prejudice to the traditional parameters. For example, the cost of breathing in moderate exercises is about 3 % to 6 % percent of total body oxygen consumption, whereas in intensive exercises it is about 10 % to 15 %. Training has been shown to correlate with improvements in the running economy by a decrease in the exercise ventilation. Therefore, a proper rhythm of breathing to SR seems to be responsible. As previously mentioned in the biomechanics section, the stride pattern impacts the oxygen uptake, see fig. 2.15, b. Training leads to improvements in the circulatory and the respiratory system but as well as to neuromuscular adaptations e.g., better muscle fiber recruitment (Karp, 2008; Bonacci *et al.*, 2009; Nummela *et al.*, 2006; Paavolainen *et al.*, 1999c,b), whereas strength training before a running session may lead to inadequate adaptions (Ho *et al.*, 2010). Since knowing this, strength and plyometric training have been added to training regimes e.g., Mikkola *et al.* (2011), in contrast to former attitude that focused on improving on $\dot{V}O_{2max}$. Magness (2009) alluded to studies that showed "improved $\dot{V}O_{2max}$ by 5 % without an improvement in performance over either 3000 or 5000 m" (p.19). Studies such as (Paavolainen *et al.*, 1999a) showed improvements in the performance without changes in $\dot{V}O_{2max}$ but with relation to the "neuromuscular capacity to produce force" and "higher pre-activation of the working muscles" (Nummela *et al.*, 2006, p.1,2). Nevertheless, on page 325 Hawley *et al.* (1997) summarized some requirements for good endurance athletes which were 1) a high $\dot{V}O_{2max}$ ($< 70\, ml\, kg^{-1}\, min^{-1}$); 2) the ability to maintain a high percentage of $\dot{V}O_{2max}$ for sustained periods; 3) a high power output or speed at the lactate threshold; 4) the ability to withstand high absolute power outputs or speeds and resist the onset of muscular fatigue; 5) an efficient/economic technique; and 6) the ability to utilize fat as a fuel during sustained exercise at high work rates.

Training methods are not within the scope of the thesis and are not discussed. Nevertheless, training is assumed to cause "chronic adaptive responses to physical training" (Midgley *et al.*, 2007, p.857). Midgley *et al.* emphasized that the documentation of training is challenging. Training load as the product of intensity, duration and frequency is missing a description for the training intensity and, therefore, has been found to be weak. Training intensity usually is based on physiological measures e.g. the heart rate, which "are all indicators of relative physiological strain" (p.859) and, therefore, poor indicators, because they omit intra-subject differences. Moreover, due to the cardiac drift during running the heart rate can increase by seven beats per minute for each loss of 1 % of body weight due to dehydration (Lambert *et al.*, 1998). Midgley *et al.* (2007) quoted the training intensity threshold as the minimal stimulus for adaptive response. Moreover, this group stated the difficulty to estimate training load during training and to plan optimal training load above the training intensity threshold. In

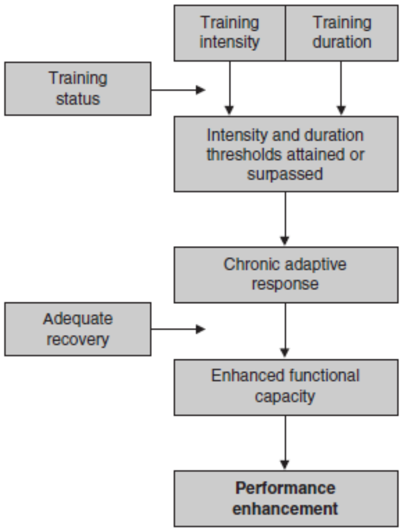


Figure 2.16: Model for the minimal training intensity threshold concept (Midgley *et al.*, 2007)

this model, it is necessary to estimate the training status and the adequate recovery. Following this model, see fig. 2.16 chronical adaptive responses lead to an enhanced functional capacity and finally to an enhanced performance. The interested reader on training methods based on the three players in performance diagnostic ($\dot{V}O_{2max}$, LT, running economy) is referred to the detailed review of Midgley *et al.* (2007) on scientific knowledge in this area. Their main conclusion was that scientific knowledge is still too little to give training recommendations, especially based on the traditional parameters. The risk of overtraining "should be an important consideration“ since enhancement in performance has to be achieved—but there were no hints as to finding out how.

The physiological view allows for profiling athletes. The physiological measures and determinants help to understand the performance of runners, and may also lead to conclusions drawn upon planned training sessions. The ambition may be warranted, because there seems to be a range of 30 to 40 % of improving performance through variations in the biomechanical and the physiological systems (Joyner & Coyle, 2008)– even if there might be neither an exclusive biomechanical or physiological variable, nor a set of these variables. For a more detailed understanding of the physiological factors, the review of Joyner & Coyle (2008) is recommended, see also fig. 2.17.

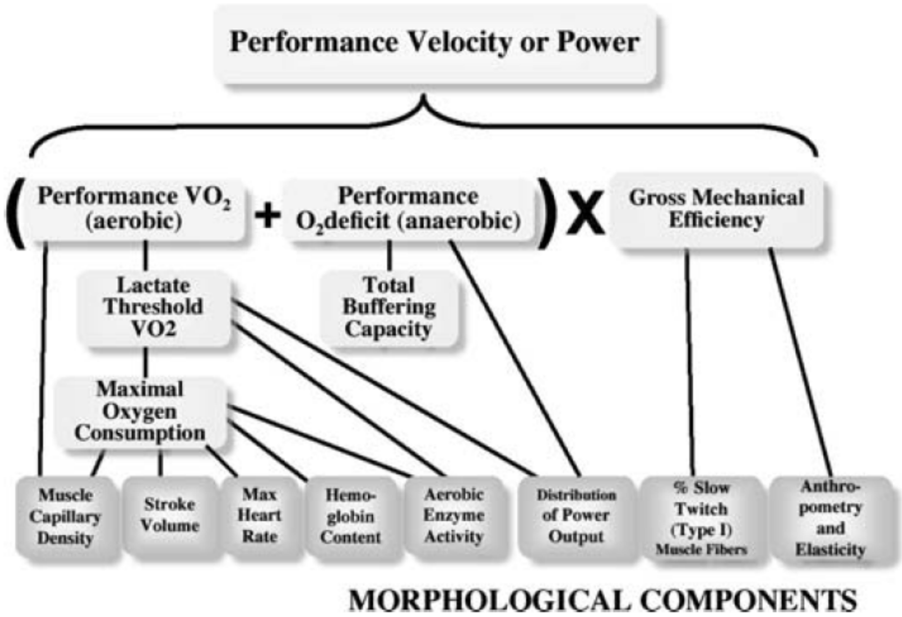


Figure 2.17: Physiological factors affecting running performance (Joyner & Coyle, 2008)

Oliver & Stembridge (2011) introduced an index combining both biomechanical and physiological measures, and moreover, which is reliable in monitoring and predicting middle-distance running. This index named heart rate-to-contact time abbreviated to *HR:1/CT* involves the heart rate and the rate of force generation, the inverse of CT and thus "reflects the integrative response of many systems" (p.432). The inspiration for such an index was reached through studies with animals or which were health-based. As earlier mentioned in the section biomechanics, CT has a central meaning to the stretch-shortening cycle. Good runners are able to use up the recoil of the muscles and tendons. Eccentric contractions are much more efficient than the concentric ones. As speed increases, CT should decrease to ensure an efficient energy return. The runner must be able to transform this energy into forward propulsion over a short period of time. This ability is associated with neuromuscular qualitites. The metabolic energy cost increases as the CT decreases, but "reduced contact times have repeatedly been shown to be associated with better economy in distance runners, suggesting a lower metabolic cost with a shorter contact time" (p.435). This is obviously a paradox. Metabolic cost increase due to the reliance on more fast-twitch muscle fibers. Well-trained runners rely more on the eccentric than on the concentric muscle activity during stance and are at least at moderate to sub-maximal speed very economic. At higher speeds, the metabolic cost increase rapidly but otherwise these speeds could not be reached, whereas untrained runners rely more on the concentric muscle activity and, therefore, do not exhibit this short CT. Fatigue causes athletes to increase CT, in order to reduce metabolic cost, but thus they also have to reduce their speed. CT representing neuromuscular fitness depends also on the individual (e.g., muscle fiber distribution), qualities in running and the overall fitness. However, the cardiovascular system represented by the heart rate, provides the runner with the metabolic energy, whereby a lower heart rate is associated with a lower oxygen consumption i.e., better runner economy. A low *HR:1/CT* index points to a non-fatigued state—resulting from a low heart rate and a low CT. For example, the index may not change although the runner is now exposing a higher speed and with that associated a lower CT. Oliver & Stembridge (2011) did not mention directly whether or not an increased index represents fatigue, because the purpose of their study was to determine the reliability of the HR:1/CT index. In accordance to their logic, a rise in the index might represent an increase in the neuromuscular stress/fatigue and/or an higher demand (due to fatigue) of the cardiovascular system.

Stirling *et al.* (2012) established another multi-factorial fatigue index calculated iteratively over equal intervals throughout an one hour run by integrating heart rate, respiration rate, SR and three psychological descriptors (strong, relaxed, energy). All the data (in each single variable) were normalized, in order to be independent of a runner-specific calibration by subtracting the mean from the current data point divided by the range of all data points. Principal component analysis was applied and "it was assumed that the greatest variation in the data was caused by fatigue" (p.3). The variance of the first component has been calculated and named 'contribution vector'. The fatigue index is the product of a current normalized data point and the current contribution vector, so that this index when considered over time can be used as a fatigue slope. The fatigue index was in agreement with the results of the Borg-Scale (Rated Perceived Extertion, *RPE*) (Borg, 1998, 1982). In contrast to the *RPE*, this multi-factorial index may allow for weighting the influences of the underlying systems. The last two studies demonstrate that there are endeavors to reveal measures that allow for evaluating the state of the athlete and predicting further performance or fatigue, respectively.

2.1.6 Injuries

"Each year between 1/4 and 1/2 of runners will sustain an injury that is severe enough to cause a change in practice or performance. This may lead the runner to seek consultation, alter training, or use medication" (Novacheck, 1998, p.77). The risk of injury cannot be neglected and has to be factored seriously when planning training and competition. This section aims to give an overview of the most common injuries and its etiologic factors. As a second goal, this section addresses the issue of the prevention of running-induced injuries. Different definitions of the terms injury and prevention have been suggested in the literature. Injury can be defined among others as a state of being "serious enough for training reduction [...] or even pain for more than 10 days" (Bangsbo & Sjogaard, 2001, p.110). According to these authors, prevention "must consider the entire complex of the kinetic chain [and] the history of individual predisposition" (p.110). Therefore, a "physical examination"(p.110) of the runners has to be undertaken but the criteria of such an an examination are still of debate—compounding to set a standard. Moreover, there are various kinds of running e.g. hurdles, sprinting, middle and long distance, orienteering, running on synthetic flat ground, crowned or asphalt roads, etc. This work pays particular attention to recreational over-ground runners on middle to long distances. Furthermore, lowering the risk of any injury is seen as a potential factor customizing healthy training and promoting lowest trade-off for long-term high performance.

Sports activities are generally considered to promote a healthy life style. Especially in the industrial countries efforts are made to "reduce the incidence of obesity, cardiovascular diseases and many other chronic health problems" (Gent *et al.*, 2007, p.469). Recreational running or jogging has been ranked highly foremost other sports in the 1990th as one of the best physical activities to prevent modern civil diseases (e.g., Dugan & Bhat (2005) and Hafstad *et al.* (2009)). Nevertheless, since the 1970s recreational and athletic running have evolved to a mass phenomenon and it has become apparent that running also causes orthopedic problems; thus Bangsbo & Sjogaard (2001) raised the question "whether or not this type of physical activity is still recommendable or even harmful". In order to approach the answer of this question, the prevalence and the etiological factors will be consulted in the following.

Running injuries occur two to two and a half times less often than all other sports. Most of the injuries in all sports are located in the lower limbs, whereby two-thirds of these injuries affect the knee and ankle joint complex (Bangsbo & Sjogaard, 2001). Compared to ball sports, in running, injuries at tendineous structures are more common. According to Bangsbo & Sjogaard (2001), the total risk is between three and half and five and a half incidences per 1000 h of running“ (p.112). Gent *et al.* (2007) revealed in their meta study that the incidence of lower extremity varies from 0.4 % to 79.3 % between studies, including all body areas the range is from 26.0 % to 92.4 %. The knee as the predominant site was numbered from 7.2 % to 50 %. The lower leg (shin, Achilles tendon, calf, heel) ranged from 9.0 % to 32.2 %, the foot from 5.7 % to 39.3 %, and the upper leg (hamstring, thigh, quadriceps) from 3.4 % to 38.1 %. Lower risk of injury was found in the ankle ranging from 3.9 % to 16.6 % and in the hip/pelvis complex ranging from 3.3 % to 11.5 %. Injuries can affect the skeletal, musculotendineous, vascular, and neurological systems as well as influenced by infectious and neoplastic processes. The most common injuries in running are the medial tibial stress syndrome, Achilles tendinopathy, tibial and fibular stress fractures and gastrocnemius/soleus strain/tears, see tab. 2.10, 2.11 and 2.12 for further injuries and descriptions. The name 'shin splint' has been widely used, and is a collective name for overuse injuries around the shin, but has been mostly associated with the medial tibial stress syndrome.

For the etiology of injuries, there is a distinction between extrinsic or environmental (surface, footwear, training, heat, cold, etc.) and intrinsic factors (age, gender, muscular system, body alignment, etc.) (Bangsbo & Sjogaard, 2001). All these factors play in concert and individual examination is necessary to determine the prevalence or the causes of injuries. However, there were studies that attempted to extract the etiological factors in the average of the runners. These results are in detail conflicting or more trends with limited evidence than significance.

Depending on the statistical analysis, it turned out that greater age is or is not associated with a higher risk of injury (Gent *et al.*, 2007). For example, multiple stepwise regression led to the conclusion that age is rather an "independent factor. Neither

Table 2.10: Running injuries - Part I

Injury	Description
Medial tibial stress syndrome	is "pain along the the medial distal two thirds of the tibia which is usually caused by tendinitis and periosteal irritation of the tibial posterior muscle and tendon, the flexor muscles and tendons, and/or other soft tissue attached to the posteromedial border of the tibia" (Messier & Pittala, 1988, p.501) (soleus). It has been associated with an imbalance of foot pressure (greater on the medial foot) and excessive pronation. Shin splint is another common used name, whereas in general it has been referred to all injuries along the shin.
Achilles tendonitis and tenosynovitis	Achilles tendon transfers the energy from the big calf muscle complex and is highly affected during high impacts such as jumping or impulsive propulsion. Symptoms develop slowly and can occur after overuse through training on crowned roads, interval training, emphasizing push-off phase or running on soft surfaces such as sand or athletic track especially when using spikes. Patient is tender to palpation or pressing at the proximal part of the tendon. Achilles injuries are in the majority (2/3) tendonitis and tenosynovitis (incidence rate 1/3). Tendonitis is characterized by irritation and swelling of the friction bearing tissues, whereas the tenosynovitis (1/3 incidence rate) causes excoriation of the bursa and painful swelling at the insertion of the Achilles tendon.
Tibial and fibular stress fractures	Stress fractures in running occur in all lower extremity bones such as tarsus, thigh, pubic bone and resulting groin pain, sacrum, but are most common in the tibial (incidence rate 50 % of fractures) and fibular bones (5 % − 21 % of fractures). Dull pain onsets during running but decays afterwards. Bones become weaker after an increase in training intensity. Bones accommodate longer than soft tissue. It can be caused by an inappropriate increase in mileage and/or an inappropriate running technique such as a too low SR compensated with a higher SL, resulting in higher shock waves traveling upwards the leg. "The mechanism of injury may be repetitive ankle plantar flexor contraction" (Gallo *et al.*, 2012, p.488). Selected further factors accord. to Gallo *et al.* (2012): low bone mineral density, lean mass in lower limbs, low fat diet in female runners, leg length discrepency, peak hip adduction, rearfoot eversion angles during stance.

Table 2.11: Running injuries - Part II

Injury	Description
Gastrocnemius/soleus strain/tears	Gastrocnemius and soleus are strained out to their complex structure. The fast-acting gastrocnemius is at the greater risk beneath both. "Injury to medial head of the gastrocnemius is caused by sudden dorsiflexion of a plantar flexed foot with the knee in extension or sudden extension of the knee with the ankle dorsiflexed. Running studies indicate that this injury occurs near touchdown and is associated with faster-than-normal running speeds and inappropriate body posture, which caues altered muscle length and shock absorption. The injury has a predilection for the poorly conditioned, middle-aged with 'thick calves' who is engaged in strenuous activity." (Gallo *et al.*, 2012, 489)
Chronic exertional compartment syndrome (CECS)	"Most commonly occurs in young adult recreational runners, elite athletes, and military recruits. CECS is caused by increased intracompartmental pressure within a fascial space" (Tucker, 2010). "During heavy exercise, fluid accumulates within the interstitial space of skeletal muscle, increasing mass up to 20 %. The buildup of interstitial fluid combined with limited expansion of the fascial compartments, especially the anterior and lateral leg compartments, may lead to elevated intramuscular pressures, causing capillary occlusion. [... Low] muscle capillary supply is a possible pathogenic factor." (Gallo *et al.*, 2012, 491) Forefoot running, an increased SR and avoiding over-striding may help to rehabilitate and prevent this injury (Diebal *et al.*, 2011).
Popliteral artery entrapment syndrome	"In the development of [this injury], the popliteal artery is focally compressed against the medial femoral condyle during forceful plantar flexion. With repeated constriction, the artery can sustain damage to its wall and forms aneurysms and/or stenotic lesions leading to thrombosis and/or embolic events. For reasons unknown, men are more prone to the disease" (Gallo *et al.*, 2012, p.492).
Iliotibial band friction syndrom (ITBS)	is "an inflammation of the iliotibial band as it passes over the lateral femoral condyle and/or inflammation at the insertion on the Gerdy's tubercle" (Messier & Pittala, 1988, p.501). Weak hip muscles, over-pronation, leg length differences, running on crowned roads have been associated with the cause of this injury. Just recently it also belongs to the injuries termed with "runner's knee".
Plantar fasciitis	is "an inlammation of the fascia, and soft connective tissue at the site of the plantar fascia attachment on the inferior aspect of the calcaneal tuberosity. In severe cases, a calcaneal spur may develop at the point of attachment"(Messier & Pittala, 1988, p.501).

Table 2.12: Running injuries - Part III

Injury	Description
Patellofemoral painsyndrom (PFPS)	has been traditionally called "Runner's Knee" since the 1970th. There is pain around or behind the kneecap. "Three major [...] factors [contribute to PFPS]: (i) malaligment of the lower extremity and/or patella; (ii) muscular imbalance of the lower extremity; and (iii) overactivity " (Thome *et al.*, 1999). The most common variation of this injury is when the patella moves too lateral during the extension of the leg. Weak medial muscles such as vastus medialis oblique and tight lateral structures such as vastus lateralis iliotibial band lateral reinaculum support the dragging away from the groove formed by the femur and tibia. Over-pronation and a high Q-angle foster PFPS. It has been often confused with chondromalacia patella.
Patellar tendinitis	has been also known as "jumper's knee" and causes pain in the patella tendon as it gets irritated and inflamed, when quadriceps contracts but also when pressing or palpating. As its name suggest it has been caused by repeated fast and high loading eccentric quadriceps contractions. It is a chronic condition and develops gradually. Weakness may be present in vastus medials obliquus and calf muscles.
Ankle sprain	has been associated with motor control deficits. After the occurrence in 30 % to 40 % accord. to Webster (2013) and 40 % to 70 % accord. to Steib *et al.* (2013) of the runners will develop chronic ankle instability originating most-likely from mechanical and functional ankle instability Steib *et al.* (2013). Especially these runners are susceptible to fatigue and high intensities to a re-occurrence.
Haglund's syndrome	"Haglund syndrome is a common cause of posterior heel pain, characterized clinically by a painful soft-tissue swelling at the level of the achilles tendon insertion. On the lateral heel radiograph the syndrome is characterized by a prominent calcaneal bursal projection, retrocalcaneal bursitis, thickening of the Achilles tendon, and a convexity of the superficial soft tissues at the level of the Achilles tendon insertion, a 'pump-bump' (Pavlov *et al.*, 1982).
Others	Extensor tendonitis, arch pain, muscle pulls, meniscopathy, blister, bloody toes.

age alone nor age associated with experience or weekly running distance is associated with injury." (Bangsbo & Sjogaard, 2001, p.112) The meta study of Gent *et al.* (2007) found four high quality studies stating a greater age as a risk factor, whereas two other high quality studies acknowledged age as a protective factor. There was limited evidence that male runners are prone to injuries in hamstrings and calf muscles, whereas female runners tend injure in the hip area (Gent *et al.*, 2007). However, in general, age and gender are no risk factors (Bangsbo & Sjogaard, 2001). There is also no significant risk for a *BMI* greater than 26 to injury. According to Gent *et al.* (2007), a greater weight may even be protective against foot injuries and a *BMI* greater than 26 would prevent the overall injury risk. Gent *et al.* (2007) and Bangsbo & Sjogaard (2001) noted that less runners with high *BMI* run high mileage or fast and, therefore, the real influence of weight on the injury risk remains shrouded. In general, the biomechanical and anthropometric structure cause 40 % of injuries (Bangsbo & Sjogaard, 2001; Gallo *et al.*, 2012). According to Bangsbo & Sjogaard (2001), those malalignments may involve: "different limb length, knee abnormalities (knock knee; bow legs; patellar deformities) or foot anomalies (varus/valgus, etc.)" (p.113). Gent *et al.* (2007) added a higher left tubercle-sulcus angle and a greater knee varus as risk factors for shin injuries. Additionally, Gallo *et al.* (2012) mentioned a greater knee varum and a higher Q-angle[1]. Moreover, males greater than $1.70\,m$ are exposed at a greater risk (Gent *et al.*, 2007).

Messier & Pittala (1988) examined the etiologic factors on three common running injuries: iliotibiales band friction syndrome, plantar fasciitis and shin splint. Shin splint was defined as "pain along the medial distal two thirds of the tibia which is usually caused by tendinitis and periosteal irritation of the tibialis posterior muscle and tendon, the flexor muscles and tendons, and other soft tissue attached to the posteromedial border of the tibia" (p.501). Definitions and description of the other two injuries can be found in tab. 2.10, 2.11 and 2.12. Leg length difference of at least $0.64\,cm$ and greater range of motion in plantar flexion was present in the plantar fasciitis group. The iliotibiales band friction group "had a slightly higher arch than the control group" (p.502). The range of motion in ankle dorsiflexion was smaller in the shin splint and

[1]The Quadrizeps-angle (Q-angle) is the angle between the rectus femoris and the patella tendon.

iliotibiales band friction syndrome group than in the control group. About 20 % of all injured subjects included hills in their training; and the same amount of the iliotibiales band friction syndrome group "ran on crowned roads" (p.503). Furthermore, the shin splint group demonstrated a greater angle and velocity of pronation. Hyperpronation is a well-known candidate inciting problems. As expected, the control group exhibited "less pronation and less total rearfoor movement, and smaller pronation velocity than all injury groups" (p.503). Links between a poor back, posterior thigh flexibility, Q-angle and the injury risk could not be established. Messier & Pittala (1988) regarded the Q-angle as being not important or considered to liberal. However, it remains unclear whether or not the correction of biomechanical abnormalities or predisposition accounting for less about the half of the injuries "will prevent or help to treat lower extremity injuries" (Gallo *et al.*, 2012, p.487), see also chapter 2.1.4.

The majority (60 %) of injuries stem from training errors (Gallo *et al.*, 2012; Gent *et al.*, 2007; Bangsbo & Sjogaard, 2001). The analysis of training related factors revealed that running a whole year through or/and longer race distance place the runner at a significant greater injury risk. This may suggest that most of the runners suffer from overuse injuries. Gent *et al.* (2007) found that running more than two times a week can cause overuse. Furthermore, their review revealed two studies, which showed that running more than 64 km per week was significantly correlated with injury risk in males. There was only one study showing this for females, but it could not be ascertained whether or not there was just one study showing this result or there were more studies but only one study was significant. Increasing the distance per week may be protective against knee injuries but a greater risk factor for hamstring related injuries, whereas increasing the duration of training per week showed some limited evidence being protective for knee and foot. Training frequency may not play an important role (Bangsbo & Sjogaard, 2001) as long as it is not increased due to a greater overall training distance (Gent *et al.*, 2007). Previous injuries are more important to a higher injury risk. In this meaning, prevention has also to care about the avoidance of "recurrence of that specific problem" (Bangsbo & Sjogaard, 2001, p.113). Corrections, i.e., shoes, changing environment, habits, mileage, etc. can be harmful, too, when they are applied too suddenly or too strongly. Even though more experienced runners can obviously better reduce the injury risk, nevertheless, they suffer from injuries—due to their higher distance and pace. Bates (2010) could show that the risk of the most common injuries "has not changed appreciably" (p.27) since the 1970th—regardless of the changes in footwear, surfaces and training methods. Even if these changes led to reduction of injury risk with respect to former level of effort, then it might be reasonable that athletes were able to train harder by increasing their mileage or intensity beyond former limits. Apparently, running injuries cannot be prevented as long as overuse is the critical factor. Overuse through high mileage or high intensity combined with an adverse running technique such as over-striding (Diebal *et al.*, 2011; Fletcher *et al.*, 2010; Romanov & Fletcher, 2007; Arendse *et al.*, 2004; Elliott & Blanksby, 1979) can lead to a higher injury risk, especially under fatiguing conditions during the training or when there is chronic overloading. Some injuries are caused by motor instability and lead to falls or ankle sprain. In fatiguing conditions and with fast pace the re-occurrence after an initial ankle sprain raises to 40 % or even to 70 % according to Webster (2013) and Steib *et al.* (2013), respectively. Such injuries happen usually within the second half of athletic events (Whiting & Zernicke, 2008). It is assumed that the motor control shows deficits (Webster, 2013). "Sensorimotor control is also temporarily impaired in physically fatigued state" (Steib *et al.*, 2013, p.1). An attenuated sensorimotor control can lead to an "increased muscle reaction time", "reduced muscular muscle activation", "altered proprioception", and altered "postural control" (Steib *et al.*, 2013, p.1), and, therefore, it reasoned that "dynamic joint stability is decreased in the fatigued state" (Steib *et al.*, 2013, p.1). The question was and is "How much is too much" (Bangsbo & Sjogaard, 2001, p114) or more in detail what is "too fast, too similar, too different" (Bates, 2010, p.29)?

Physical training yields environmental "demands/stimuli" causing the anatomical structures to modify (Bates, 2010, p.29). If sufficient load has been applied to these structures, then a physiological response can lead to accommodations i.e., a positive change increases the acute threshold of the physical load. If the load on these structures "exceeds the tissue's physiological ability to accommodate, then it will be become damaged" (Bates, 2010, p.29) and the injury risk increases. The etiological factors all do have an influence on the instant load. For example, Verbitsky *et al.* (1998) demonstrated that "fatigue hampers the ability of the human musculoskeletal system to protect itself from over-loading due to heel strike-generated shock waves." (p.301), see next section for details. Variability as previously mentioned is inherent in all movement. In the light of the injury risk, it supports avoiding chronic overloading. Running offers a variety of possibilities to accomplish the movement task of the cyclic movement. A certain amount of variability has to be reduced to save energy, but some variability is necessary to prevent overloading. As it has been mentioned previously, variability might change over the time course and especially during fatigue. From the literature, it did not become apparent, whether or not the amount of variability decreases during fatigue while Meardon *et al.* (2011) gave hints that the structure of variability might be a predictor for the injury risk in over-ground running, see chap. 2.2.

2.1.7 Biomechanical adjustments over time

It could have been substantiated that a complex interplay between several factors has an influence on performance as well as on injury risk. This section is concerned with the time course of the stride parameters and attempts to reveal possible adjustments and their relation to performance, the risk of injury and to fatigue, which especially in time-trials, progresses and seems to interconnect both the performance and the risk of injury. To the knowledge of the author, there were only few studies investigating stride kinematics in middle-distance over-ground running, therefore this review includes studies using treadmill and/or long-distances, too, in order to extend the theoretical background for prospective hypothesizing.

Treadmills have been chosen for mainly two reasons over over-ground running. First, the running speed can be controlled and thus adjustments can be ascribed clearly to fatigue-related accommodations, hence behavioral changes can be excluded (Siler & Martin, 1991). Second, stride kinematics can be captured with a high accuracy and precision for each step. For example, video-based (Hardin *et al.*, 2004), accelerometer-based analyses (Derrick *et al.*, 2002) and kinematic arms with e.g., opto-electronic markers (Candau *et al.*, 1998; Belli *et al.*, 1992) have been used for this purpose, see chap. 2.3 for more details on measurement devices. Stride kinematics differ in over-ground and treadmill running from each other. In over-ground running, SL is longer and SR is slower (Schornstein, 2011) while the knee angle, an important factor for the leg stiffness, displays a greater maximum and a smaller minimum.

Brueggeman (2009) stated that lower leg kinematics change "in different stages of endurance activity, presumably as a result of fatigue" (p.29). Muscular (local) and neural (central) fatigue are responsible for a diminished motor potential. Muscle activation pattern can no more efficiently react to stretch loads. This leads then to a reduction in leg stiffness. As a reaction at initial contact, the knee angle increases and "the subtalar joint becomes more inverted" (Hardin *et al.*, 2004), rearfoot angles i.e., pronation increase (Brueggeman, 2009); this is accompanied by peak accelerations of until 50 % increase (Candau *et al.*, 1998)—and has been associated with a higher risk of injury (Verbitsky *et al.*, 1998). But if this increased acceleration is due to a decrease in the effective mass than this compensation can be seen as a strategy to avoid fatigue-related injuries, because then at the same time the impact forces decrease. This compensation is expensive in terms of metabolic cost and thus translates into a decrease in performance. For example, flexing the knee (in order to stretch the leg extensors and therfore increase the leg stiffness) by 5 ° more than usual equals 25 % more energy (Hardin *et al.*, 2004). It is of debate "whether these changes in kinematics were the result of a strategy to shift the optimizing criteria from performance to injury prevention or it was a failure of the system to maintain optimal behavior" (p.1002). Fatigue causes the system to slow down; and dependent on the cause of fatigue, strategies for increasing performance and/or avoiding injuries have been discussed. If fatigue is mainly caused by a failure in the neuromuscular mechanism, then training such as plyometrics can be suggested (Ho *et al.*, 2010). In case of being caused by injury, prevention then perhaps footwear might handle the impacts (Hardin *et al.*, 2004). However, regarding injury (chap. 2.1.6), it has been demonstrated that high impact forces as well are not the only reason for higher risk of injury, rather it depends much more on the ability of the muscle-tendon complex to attenuate the shocks and thus soft-tissue vibrationssee chap. 2.1.4. The author could not find hints in the study of Hardin *et al.* (2004) as to whether or not the kinematic adjustments necessarily lead to a reduced effective mass and, therefore reduced impact forces. Furthermore, these investigators confirmed the inverse relationship between effective mass between peak impact forces and acceleration. According to Brueggeman (2009), the increased cost during fatigue might also be caused by "undesired accessory forces perpendicular to the defined task" (p.35). He also recognized that there was a progress in fatigue in aerobic and anaerobic exercises, but being strongest in the anaerobic exercises. At first, fatigue evolves in the slow-twitch fibers, and then in the fast-twitched fibers. Steib *et al.* (2013) speculated that the central rather than local control mechanism is disturbed by fatigue. Multiple sensory information is taken into account for the postural stability. As an interpretation by the author, this could mean that other factors such as poor mental state and/or high cognitive loads can disturb this neuro-muscular regulation. If runners are in a healthy physical and mental state there is only little day-to-day intersubject variability in the kinetics and kinematics of treadmill running (Morgan *et al.*, 1991). This is suggestive of the assumption that a runner maintain one's running pattern (including adjustments) over different trials as long as environmental conditions are controlled.

For middle-distance over-ground running, Saziorski *et al.* (1987) proposed a model for elite runners predicting the time-course of the stride parameters within three phases of nearly the same duration. The first phase shows no signs of fatigue but a stable relation between SL and SR. In phase two, SL decreases, but SR compensates and, therefore, the stride speed (SS) is nearly the same as in phase one. This neuro-muscular adaptation happens unconsciously. Finally, both parameters decrease, and so does SS, whereas the overall coefficient of variation in SS is between one and three percent. Unfortunately, there was little evidence presented as to the model's validity. (Elliott & Roberts, 1980) confirmed parts of the predictions. Over the time-course of a $3000\,m$ time-trial, SL decreased, whereas SR increased. The increase in contact-time (CT) and the decrease in flight-time (FT) were also in consensus with the theory. As it expected, the knee flexion at initial contact increased. These significant changes occurred mainly in the last of four stages being equal in length. The eight college runners had a low variability in overall speed. In the study of Nummela *et al.* (2006), eighteen well-trained competitive runners showed a decrease in CT and FT during a $5\,km$ over-ground time-trial. In contrast, Verbitsky *et al.* (1998) experienced a decrease in SR during an all-out run at the anaerobic threshold velocity (trial lasted about $30\,min$). The well-trained, but not competitive male runners also showed this decrease in SR when this parameter was normalized to speed. Results of SL were not presented. Competitive female and male runners of the study of Hanley & Smith (2009) were slightly different in their adaptations, but displayed an overall decrease in both SL and SR. Stride kinematics seem to behave differently on treadmills. The main difference is the increase in SL and the decrease in SR—opposite to the expectation in over-ground running. The ten recreational runners of the study of Derrick *et al.* (2002) ran at their 3200 m pace until full-exhaustion. At initial contact, high peak acceleration and higher knee flexion and a higher dorsal flexion have been detected. In the study of Mizrahi *et al.* (2000), 14 male recreational runners exposed a decrease in SR and an increase in knee flexion. Data for SR was not given, though due to the constant speed of the treadmill belt it can be speculated that SL increased. During fatigue, hip excursion increased; therefore, the SL should not have been increased necessarily but the vertical oscillation of the center of mass. It might be that treadmill running causes special modifications in kinematics. The 27 recreational runners ran at 70 % of their VO_{2max}. The test lasted 30 min. They exposed an increase in SL and step width—that must have resulted in a lower SR. Competitive triathletes were asked by Candau *et al.* (1998) to run 3000 m on a treadmill. The 15 athletes showed a slight decrease in step rate; its variability also increased slightly. These changes were accompanied by higher (metabolic) cost of running. Although there were no words about SL, due to the constant speed of the treadmill belt, it should have increased slightly.

Landers *et al.* (2011) examined triathletes during a competition, running 10 km as the last of the three disciplines. The overall SL decreased while SR remained stable. In the study of Kyroelaeinen *et al.* (2001), seven male and six female triathletes ran a Marathon at an individual constant pace, which was determined in a sub-maximal running test on a treadmill. Over the whole run, SL decreased, whereas SR increased. The same time-course was detected in forty female Marathon runners (Buckalew *et al.*, 1985). Similarly, SL, SR, and the running speed decreased while CT increased as Chan-Roper *et al.* (2012) observed competitive runners in a Marathon. When SL, SR and CT were normalized for speed than these stride parameters increased by 1.3 %, 2.0 % and 13.1 %, respectively. Furthermore, they found no difference between the compensations of the faster and the slower runners. Hunter & Smith (2007) let their trained athletes (with a training regimen of at least 32 km per week) run at a range from 96 % to 99 % of their 10 km racing speed for one hour. SR decreased and, therefore, it is supposed that SL did increase. Dutto & Smith (2002) undertook a similar study and found a decrease in SR. Ten kilometers running on a treadmill revealed in good and poorer runners participating in the study of Siler & Martin (1991) that SR could be considered as constant while SL decreased and CT slightly increased. Another kinematic compensation was found in the study of Hanley & Mohan (2006) with competitive athletes running at 103 % of their personal 10 km best time. Both parameters, SL and SR, remained stable over the 10 km treadmill run. CT decreased as FT increased. Morin *et al.* (2011b) investigated the development of stride kinematics in a 24 h run on a treadmill. Ten healthy men decreased their SL and increased their SR during the run. CT decreased while FT remained stable.

Adjustments were apparent in nearly all studies, but differed between running distances and between treadmill and over-ground running. Most of the adjustments were apparently a compensatory strategy of vertical stiffness through leg stiffness—changing the effective mass through changed angles in the knee, hip, foot and overall posture. Over-ground running in the middle-distance races pointed to a decrease in SL, an increase in SR and an increase in CT. SL was directly linked to performance, whereas SR and mainly its increased variability might have caused higher cost of running. Better runners demonstrated a more stable behavior of SR and longer SL. Poorer runners tended to over-stride (Siler & Martin, 1991); this caused a faster depletion in energy; moreover, compensation through SR stresses the neuro-muscular regulation due to temporally inappropriate leg stiffness and induced higher cost of running. Finally, the overall performance deteriorated and/or risk of injury might have increased. Several studies stated that the recognition of changes in the stride parameter during the run might help to train stable running patterns. It is noteworthy that there were no differences between runners of different technique levels with respect to the structure, but rather differences were attributed to the temporal occurrence of the compensation mechanism. Kinematic adjustments are involved in the continuous optimization, whereby two strategies have been favored—the optimization towards a minimum energy overall expenditure and shifting all subsystems towards avoiding overuse or injury by restraining the overall performance. The author concludes that the strategy of minimizing the energy expenditure might be preferred especially in early stages of a run, whereas avoiding overuse does in later stages of the run. Fatigue might emerge as both crossover.

2.1.8 Summary

Performance in (middle-distance) running depends on several factors while none of them can be identified as the major variable. Traditionally, runners aim at improving their physiological performance and focusing on a proper running technique. It is known that strengthening and stabilizing training improves posture and the timing of the muscle contractions as well as the stretch-shortening cycle. Most of the training recommendation focus on the "good" running form that should be kept as long as possible. The deterioration of technique during the run is known to start anytime, but particularly its acute and long-term effect cannot be measured. Neural-muscular and especially central-nervous system's recovery last longer than muscles can refill their energy stores and can be underestimated by the athlete. Physiological measures such as heart rate variability can provide information about the stress level but less about a reduced motor potential. Obviously, there are phases, in which adjustments such as in kinematics take place. In training, the coach has the possibility for a visual inspection; and the runner also has own perceptions about changes. Incalza (2007) asked "Are we absolutely certain that an athlete is able, always and in any circumstance, to match his/her motor potential to the technical and mechanical parameters of endurance running in an automatic and instinctive manner“ (p. 42)? A biomechanical assessment involving the stride parameter could support better profiling of athletes and provide a more detailed knowledge of their performance. Another point emerging with the parameter SR seems to be the rhythm of the movement (Incalza, 2007). From the kinematic adjustments alone, one can speculate that an elevated variability in SR might have caused disturbances. Most of these slight changes would require an appropriate measurement tool and also a sophisticated analysis method in order to detect perturbed movements, which might indicate a higher risk of injury (Meardon *et al.*, 2011), see chap. 2.2. A biomechanical assessment might be justified as it intends the correct training errors responsible for the majority of injuries. Fig. 2.18 summarizes the main factors being important to the thesis, while the neuromuscular capacity has central meaning to the biomechanical approach of the MC system.

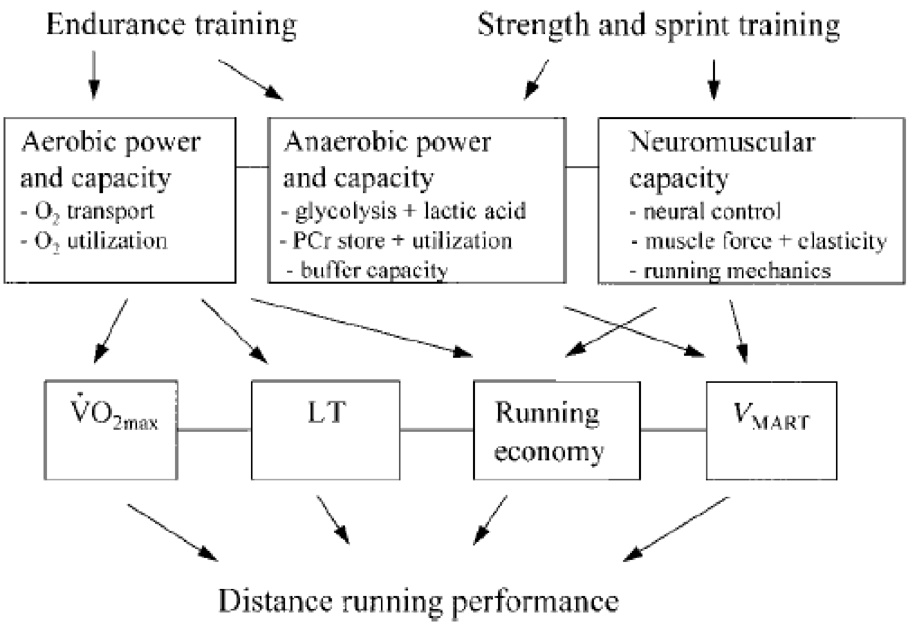


Figure 2.18: Model of determinants in distance running performance accord. to Paavolainen *et al.* (1999b) In comparison to other introduced models this depicts the neuromuscular capacity, which has a central meaning to the thesis' approach of the MC system.

2.2 Variability and stability in stride kinematics

A traditional approach to biomechanical assessment of human locomotion such as running is an analysis aiming at the "identification of invariant properties" (Bartlett *et al.*, 2007, p.233). Furthermore, it has been assumed that there is an "optimal motor pattern or movement technique" (Bartlett, 2004). Intra-individual or intra-trial, but especially variability during cyclic movement has been widely referred to "noise, not an important issue in research design or measurement" (Bartlett *et al.*, 2007, p.224). This "noise" has been associated with a complex interplay of the subsystems involved in the movement. Small variability appeared to be preferable; though the better athlete has the ability to reduce the degree of freedom and thus avoids unnecessary movements, which would be inefficient. For example, in this view, the motor control system of a runner might adjust one's kinetic chain to an average near its optimum (depends on e.g., running abilities, age or fitness). The deviation from this target would then lead to higher costs of running. However, even the best performers of their discipline show remarkable intra-individual and intra-trial variability. "Increasing expertise does not lead to movement invariance and the construction of a single, pre-determined motor pattern" (Davids *et al.*, 2003, p.250). In contrast to the traditional school of thought, movement variability could be ascribed to having a functional meaning. Movement variability can "provide a broader distribution of stress among different tissues, potentially reducing the cumulative load on internal structures of the body" (Bartlett *et al.*, 2007, p.234). For example, though a runner might be able to reduce high impact shocks. Such a functionality led to the "variability-overuse injury hypothesis" (Bartlett *et al.*, 2007, p.234), see also fig. 2.19. Furthermore, variability seems to be sufficient to adapt to changes in the environment and thus enhance performance. There have been a variety of models of motor control strategies. For example, motor task would evoke motor programs according to the impulse-time theory. These motor programs are invariant in their structure but adjustable with variant parameters in time and intensity (Schmidt, 1975). However, recent research disagrees with those thoughts. The replication of an optimal movement pattern would be bound to only few constellation, whereas sports requires fast adaptations to new situations (Glazier *et al.*, 2006) e.g., by compensating disturbing influence. In this way, variability can stabilize movements—carrying redundancy and thus equifinality i.e. "there are many, possibly an infinite number, of ways to

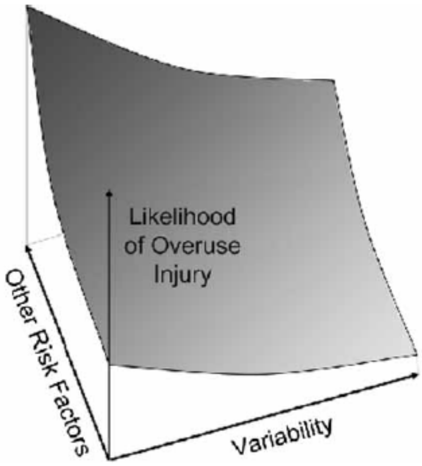


Figure 2.19: Variability-overuse injury hypothesis (Wheat (1985) in Bartlett *et al.* (2007))

perform the same task" (Cusumano & Dingwell, 2013, 3). Thus far, it has been widely accepted that variability is inherent in all biological systems, hence in the sensorimotor system and can have functional meaning within and between the underlying systems. Contrarily, variability as a process of adaptation has not been fully substantiated. It is not yet clear, whether or not the neuromotor systems exploits redundancy or attempts to overcome variability. For human treadmill walking, Dingwell *et al.* (2010) found that variability i.e., fluctuations in SL and SR, is organized by a governing principle of the neuromotor system. Stride-to-stride fluctuations are herein regulated "independently, but in parallel with, the principle of minimizing energy cost" (p.8). This study further increases confidence in the analysis of variability in gait analysis.

2.2.1 Empirical gait measurements

Chau *et al.* (2005) gave a comprehensive review of the analysis of kinematic variability in gait data. Their intention was to suggest robust estimators, whereas the mean and the standard deviation (STD) are "highly susceptible to small quantities of contaminant data" (p.4) arriving from e.g., measurement devices. They suggest to use the coefficient of variation (CV), the interquartile range (IQR) of the sample between the 75 % and 25 % quantiles, and the median absolute deviation (MAD), which is the median of the "absolute difference between the sample values and their median value" (p.4). To the knowledge of the author, there was only one study concerned with the effects of fatigue on stride kinematics during a 5 km over-ground running time-trial. Hanley & Smith (2009) were interested in how selected kinematic variables would change, especially between men and women. Variability was given as part of the standard descriptive statistics; though they numbered the average STD for three consecutive steps and three rounds. These values differed between men and women, all competitive runners. In the men, STD increased slightly, whereas in the women remained nearly constant over the first laps, then decreased slightly. It was the other way round for SR in the men. There it decreased while SL increased. Nevertheless, Hanley and colleges did not interpret the variability, because it was not within the purpose of their study. It is also questionable whether or not three steps might be sufficient to establish a robust measure of variability. In treadmill running, Belli *et al.* (1995) found that 32 to 64 consecutive steps are necessary to study intra-individual step variability of the vertical displacement of the body and step time. In contrast, Owings & Grabiner (2003) stated that at least 400 steps are required for an accurate estimation of the step variability when a high statistical power and significance level of 0.05 is assumed. The study of Morgan *et al.* (1991) revealed that the between-trial variability is little and, therefore, only few measurements are needed (if conditions are controlled). From these studies, it appears that individual movement patterns are quite stable and, therefore, potential adjustment can be ascribed to special conditions under test such as fatigue or any pathology. For example, the risk of falling during walking could be associated with a higher variability in the stride time. In the elderly population of the study of Hausdorff (2005), fallers illustrated a higher STD and CV than non-fallers (STD: 18 ms to 58 ms, CV: 1.7 % to 5.3 %).

Terrier & Schutz (2003) applied the walking relation and the stride rate index, in order to analyze the "variability of gait patterns during unconstrained walking"(p.554). The walking relation (WR), see eq. 2.17, "provides an indication of the spatio-temporal adaptation of gait at a given speed" (p.555), whereas the stride length index (SLI) "assess the relative contribution of SL and SR to the change in walking velocity" (p.555), see eq. 2.18. Alternatively, the stride rate index (SRI) can be used by replacing SL by SR in its equation. In walking, WR does not change over a large range of speed. This relation has not yet been applied to running. There is no information available about the behavior of this relation in running under fatigue. As a conclusion from chap. 2.1.7, SL might decrease, the compensatory function is suggested but only few studies could prove this; nonetheless, it can be expected that the running relation (RL) would shift to a greater portion of SR—in theory at least in the second, the compensatory stage of fatigue. In walking, SR is more responsible for the main increase in speed than SL, between slow and preferred walking speeds. Both SL and SR contributed similarly to higher walking speeds. In middle-distance running, SL appears to be the performance indicator, and under non-fatigue conditions SL has the highest influences on the speed. After the onset of fatigue, it is expected that SR would compensate partly, because due to the increased variability in SR, as well compensating for the changed leg stiffness; SLI is expected to decrease. In middle-distance running, it was of debate whether or not the variability would increase with fatigue and whether freely-chosen or a constant pace would have a favorable effect on the metabolic cost. According to Cottin *et al.* (2002), it turned out that there is no difference in oxygen consumption between constant or variable pace (the CV was 5 %); the slow phase component is not influenced. However, a constant pace has been suggested. Variability in stride speed did not increase. Under fatigue, at critical speed and in middle-distance the running speed can be quite stably adjusted, and is not the criteria for the mechanical efficiency, rather it is the goal. For an efficient run, it is expected that SL would display a remarkable amount of variability if there is variability in the stride velocity, because SR is supposed to be (more) stable.

$$WR = \frac{SL}{SR} \qquad WR \ldots \text{walking relation} \tag{2.17}$$

$$SLI = SL \frac{\log \frac{SF_{i+1}}{SF_i}}{\log \frac{SS_{i+1}}{SS_i}} \tag{2.18}$$

$SS \ldots$ stride speed, $\quad i \ldots$ ith stride in time series

2.2.2 Non-linear dynamic approaches

In the previous section, the functionality of variability emerged through the amount of the variance in the kinematic data. This amount cannot only be referred to (white) noise. Classical statistical methods are able to enlighten some of the aspects in motor variability. Although a certain amount of variability may assure adaptability, descriptive statistics works on the average of repeated movements and, therefore, cannot explain or describe the inner structure of the movements, and is not able to draw conclusion about the stability of the movements. Moreover, the amount of variability is often so small, that changes remain undetected by e.g., statistical significance. It is difficult to number a appropriate amount of variability precisely (Piek, 1998). Nevertheless, an athlete can ensure high performance when being able to generate stable movements and as well adapt motor output quickly to new demands. This has been received as the paradoxical relationship between stability and variability. In this very general view, variability helps to identify and adapts to constraints of the motor task, the environment (Glaziera *et al.*, 2003), and the person (Davids *et al.*, 2003).

There is a variety of methods to analyze gait data e.g., neuronal networks (Fischer *et al.*, 2011; Janssen *et al.*, 2011; Lamb *et al.*, 2011), support vector machines (Fukuchi *et al.*, 2011; Begg & Kamruzzaman, 2005; Lau *et al.*, 2008), wavelet methods (Lakany, 2008), and fuzzy (O'Malley *et al.*, 1997; Loslever & Bouilland, 1999). A comprehensive review is given by Chau (2001a), and Chau (2001b). In general, these methods are used to find optimal and therefrom deviating movements (states) and have mostly descriptive function. Although some of them include the analysis of movement variability, there is no direct connection of variability with stability or they lack of functional explanation of it. However, for example, information theory (Davids *et al.*, 2006) and the sensorimotor synchronization paradigm (Aschersleben, 2002; Repp, 2005) can deal with (motor) redundancy and its role in the error-correcting process. Nevertheless, according to Cusumano & Dingwell (2013), there are four prominent frameworks for the analysis of movement fluctuations , emphasizing respectively, 1) goal equivalence and task manifolds, 2) stochastic optimal control, 3) local dynamic stability or 4) fractal dynamics. The former framework assumes that the result of an action is known, solid and minimal in its cost. Due to this goal equivalence, all trials are evaluated with respect to the solutions in the task space. The corrections of the motor system are analyzed by relating the statistics of the task solutions of several trials to the task manifold (corresponding to a perfect task execution) calculated by numerical simulation. An extension accounts also for the costs between adjusting task solutions and acceptable performance. These costs determine the effect of small-body (e.g., a joint angle) and overall-body fluctuations on achieving performance nearest the goal. This framework has been impelled to a stochastic optimal control method, which is also a computational framework. Compared with framework one, it is not only data driven and, therefore, allows explanation and prediction of how the motor system "regulates variability in ways that maximizes task performance while minimizing control effort and allowing for adaptability" (Cusumano & Dingwell, 2013, p.4). Their strength is to "examine the effect of motor redundancy" (Cusumano & Dingwell, 2013, p.7). The former two frameworks can be assigned to the motor approach i.e., they are based on a hierarchic organization of the control structures, a central representation of the movement, and a representation of error and noise.

However, the action approach includes dynamical system theory based methods. For example, the local dynamic stability method attempts to assess, how the system "responds to sufficiently small perturbations" (Cusumano & Dingwell, 2013, p.4) from external sources e.g., the walking surface, and internal sources such as the neuromuscular system. Therefore, for example, the Lyapunov exponents characterize the convergence/divergence of neighboring trajectories (of e.g, positions or angles) over time (Jordan *et al.*, 2009). The maximal Lyapunov exponent is the most famous one and indicates the local stability (small values are referred to high stability). If the systems behavior is periodic, then the Floquet multipliers can be applied. It can evaluate the orbital stability with the help of Poincare maps. This allows to determine the rates "at which small perturbations away from the limit cycle grow or decay" (Cusumano & Dingwell, 2013, p.5). In contrast to the Lyapunov exponents method, the orbital stability can give evidence for the stability considering a longer range across strides. The human gait is rhythmical, but not strongly periodical. Lyapunov exponents method stems from chaos theory and assumes that there is a chaotic behavior in the movement. It has not been fully validated that these assumption are satisfied. However, both methods have been applied (with adaptations to the infringed assumptions) to (treadmill) walking gait (Terrier & Deriaz, 2011; Bruijn *et al.*, 2009; Dingwell & Kang, 2007; Dingwell *et al.*, 2001; Dingwell & Cusumano, 2000).

This approach can also be extended with a nomothetic perspective. Herein, repeated movements are established with the help of a coordination pattern, and finally a movement pattern—that is different from an optimal movement. A pattern derives from self-organization (Glaziera *et al.*, 2003) while coordinate structures temporarily join together. Due to this assemblage, the degree of freedom has been reduced dramatically. Within this self-organizing process ”preferred coordination or ’attractor’-states have been developed to support goal-directed actions“ (Glaziera *et al.*, 2003). This approach shows that the variability originates from sliding between these attractors; and movement is most stable in the center of such an attractor (Jordan *et al.*, 2007b). During this self-organizing process, the current movement is the start point of the next repetition of the movement—thus a repetition is influenced by the previous ones. Approaches based on these assumptions are embedded in the non-linear dynamical system theory that looks not at the magnitude of variability but rather at its dynamics. The study of Gates *et al.* (2007) is representative of the mechanistic approach to $1/f^{\beta}$-noise, demonstrated that fluctuations behaving in a fractal-like manner can be generated by a pure simple mechanical model incorporating sensory and motor noise. Regarding their results, the non-linear behavior of fluctuation do not necessarily stem from the neuromotor system, especially the nervous system.

In exploring the dynamics of the locomotor system's variability, it has been observed that it acts according to a fundamental phenomenon found in statistical physics—the fractals. This $1/f^{\beta}$-noise is seen to occur ”across a range of different systems and behavior“ (Torre & Wagenmakers, 2009) while not only the nervous system incorporates self-organizing characteristics. Physiologic systems are characterized by an extraordinary complexity, nonstationary and non-linear signals (Goldberger, 2002). Herein, physiologic time series are considered as a self-affine process with self-similar structures. Such a process follows a power-law function, ”which is the only mathematical function without a characteristic scale“ and, therefore, is ”called scale-free“ (Hardstone *et al.*, 2012, p.2). In theory of the fractals it is possible to zoom in or out to any factor of the used scale, nevertheless, the same structure i.e., fractal would appear. Another hallmark of fractals is that size and frequency of their fluctuations are inversely related i.e, small fluctuation have high frequencies and vice versa (Jordan *et al.*, 2009). The fractal-like behavior of repeated movement has been investigated in the majority of gait analyses with the detrended fluctuation analysis (*DFA*) and a power spectral analysis for fractals. The (relative) dispersion analysis, similar to the *DFA*, has been used, too (Chau, 2001a). For a full overview of the different approaches to movement variability and stability the interested reader is referred to e.g., Torre & Wagenmakers (2009), Bernstein (1967), Davids *et al.* (2006), Piek (1998), and (Kelso, 1995).

Physiologic data is often nonstationary i.e., they may show time-dependent trends, which veil its fractal characteristics. Analyzing nonstationary signals with conventional methods could detect spurious long-range correlations (Peng *et al.*, 1995). The *DFA* vanquishes this problem by removing local trends and, therefore, it is relatively robust against nonstationarities (Hausdorff *et al.*, 1995). Moreover, it reduces noise effects with relative simple methods.

Finally, in case of a linear relation between $\log F(n)$ and $\log n$, the *DFA* determines the slope, the scaling exponent α. This exponent can be considered as an equivalent of the Hurst exponent, which indicates a long-range correlation between the fluctuations over wide ranges, herein, whether or not "the fluctuations in the small boxes are related to the fluctuations in larger boxes in a power-law fashion" (Goldberger, 2002), and expressed as a formula:

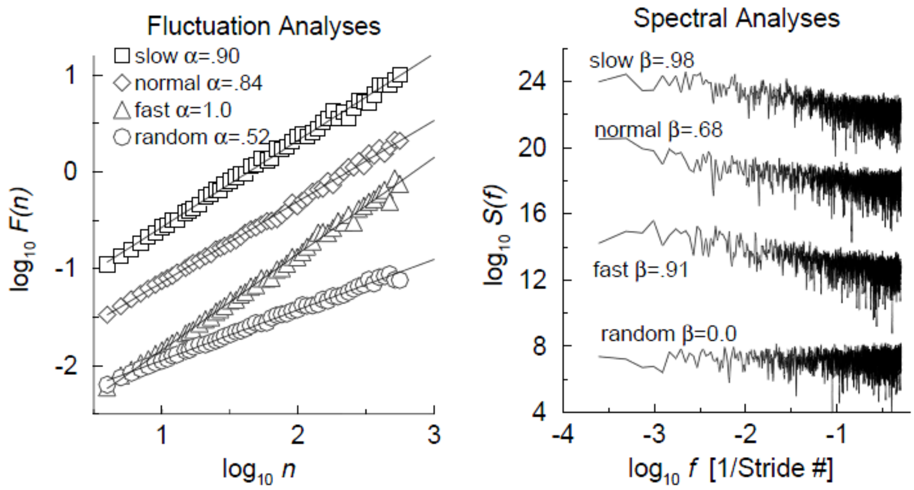
$$F(n) \sim n^{\alpha} \tag{2.19}$$

In general, values of α ranging from 0 to 1 are part of fractional (and fractal) Brownian motion. If α is 0.5, then the fluctuations are randomly distributed; they show white noise. Below 0.5 fluctuations are anti-persistently correlated. Brownian noise (smoother than white noise) is at 1.5, there is no correlation between the fluctuations. A fractal-like behavior is indicated in between white or Brownian noise while the fractal behavior $1/f^{\beta}$, also called pink noise, is perfect at unity. A random walk displays Brownian noise over short time scales, whereas over larger time scales it shows white noise (Goldberger, 2002). With the *DFA*, only one fractal property can be estimated; it is a mono-fractal analysis, see fig. 2.20, a. There is evidence that physiologic signals contain multi-fractal properties—requiring just as much exponents (Goldberger, 2002). The *DFA* has also been developed further to the multi-fractional *DFA* (Kantelhardt *et al.*, 2002).

In order to examine the presence of a long-range correlation, a log-log power spectral analysis reveals similar results as the *DFA* does—as long as the signal is stationary. Then, the time series of the signal has to be decomposed into its sine and cosine waves. The squared amplitudes of the waves will be denoted on the ordinate axis, the frequencies on the abscissa. If this time series has fractal properties, then low frequencies waves will have high amplitudes, and vice versa (Torre & Wagenmakers, 2009). The power-law emerges here as well by that decaying of the amplitudes with higher frequencies, as formula:

$$P(f) \sim f^{-\beta} \tag{2.20}$$

A linear regression determines the slope, the exponent β, which estimates the long-range correlation, see fig. 2.20, b. The exponent β is $2\alpha - 1$ (Hausdorff *et al.*, 2001). To a smaller degree than the spectral analysis, the autocorrelation functions may display long-range correlations, too, but do not have the ability to characterize the fractal behavior. Dynamical analysis has been applied to a wide range of scientific fields. "Its power lies in its generality" (Rapp, 1994, p.311), and that causes a "potential for misapplication"(p.311). For example, using the *DFA* is also prone to false interpretation. The exponent α is not a fractal dimension; it has a relation to the Hurst exponent, but does not share its properties. "Simple changes in the amplitude of sensory and/or motor noise" (Dingwell & Cusumano, 2010, p.349) can lead to an α indicating fractal-like behavior. Dingwell & Cusumano (2010) and Rapp (1994) recommended to apply the analysis methods also to different kinds of surrogate data. If the real data suggest the presence a non-linear (fractal-like) behavior, whereas the surrogate data contain no long-range correlations, then the presence of the non-linearity has been corroborated. The interpretation of the non-linear behavior can be challenging, too. Is a loss in the persistence of long-range correlation associated with a degrading of the motorsystem abilities? For example, walking under metronome condition leads to white noise in the stride interval or even to an anti-persistence correlation (Terrier *et al.*, 2005). Goldberger (2002) associated the diminished correlation with a complexity loss i.e., deterioration of the adaptability of the motorsystem. According to Dingwell & Cusumano (2010), this behavior should not be referred to a pathological mechanism, but it might even represent an "enhanced supraspinal control" (p.349). The anti-persistence correlation might be due to the fast corrections of the motorsystem with respect to the metronome. The author concludes from their results, that the overall movement might remain stable, because instead of an inner oscillator or stabilizer, the metronome has



(a) (b)

Figure 2.20: (a) *DFA* (b) Power spectral analysis (Hausdorff *et al.*, 2001)

now overtaken this function. Thus the motorsystem might thus minimize cost for stabilizing the movement. On the other hand side, it has been shown that a variety of diseases, injuries, and advanced age lead to a reduction in long-range correlation (Goldberger, 2002). Dingwell and Cusumano concluded that both a decreased and an increased control can cause the diminished correlations.

There is a large body of investigations demonstrating that the human gait of walking behaves in a non-linear, dynamic, and even multi-fractal manner (West & Latka, 2005), see e.g., Terrier & Deriaz (2011), Hausdorff *et al.* (1995), Hausdorff *et al.* (2001), Terrier *et al.* (2005), Goldberger (2002), and Dingwell & Cusumano (2000). Most of the studies concentrated on the fluctuations in the stride time (ST) of the walking gait. These fluctuations exhibited long-range correlations while factors such as disease, fitness, and age impinged on the strength of their correlations i.e, on the stability of the movement (Hausdorff, 2007; Goldberger, 2002; Hausdorff *et al.*, 2001). Hausdorff *et al.* (2001) demonstrated that subjects with the Huntington's disease could be separated from their healthy counterparts through a lower exponent α of ST in the walking gait. In general, elderly exposed fluctuations closer to white noise, whereas younger subjects exposed fluctuations more in the middle of Brownian and white noise. The influences of speed on the stability emerged as CV increased slightly, whereas the exponent α of both SL and ST appeared in a U-shaped manner with the minimum (approx. 0.74) at the preferred walking speed (Jordan *et al.*, 2007b). In a similar study, Jordan *et al.* (2009) showed that the stability at the transition speed from walking to running is decreased (decreased local dynamic stability and autocorrelation). A stable movement was characterized with a high local dynamic stability, persistence autocorrelation and relative low kinematic variability. In the study of England & Granata (2007) the kinematic variability displayed an U-shape behavior with respect to the walking speed (mimimum was also near at the comfortable walking velocity). However, fewer studies have been undertaken on human treadmill running e.g., Nakayama *et al.* (2010), especially on over-ground running e.g., Meardon *et al.* (2011). The findings of Jordan *et al.* (2006) in treadmill running were, that there is also an U-shaped function of the stride interval with α equaling approx. 0.74. Overall, the exponent α was a little bit weaker than for walking as in most others studies. From 80 % to 120 % of the preferred running speed, CV decreased slightly in ST and SL. In a similar study of Nakayama *et al.* (2010), CV was approx. 1 % for the trained runners over the whole range of speeds, whereas CV increased with speed in the non-runners. The exponent α increased slightly from approx. 0.7 to approx. 0.8, whereas in non-runners was relative stable, but higher (approx. 0.9). From their results, it appears that variability is better controlled—its amount is relative small while its structure displays long-range correlation, which is not fully predictable—in the trained runners. Nonetheless, the effect of training has its impact only on the amount of both kinds of variability through higher speed "but not due to a difference in general running strategy involving the selection of an average stride interval (and probably step length) over a given range of running speed" (p.334). Meardon *et al.* (2011) compared ST and its variability of injured and non-injured runners in a full-exhaustive middle-distance run. In both groups, ST and its STD did not change over time, whereas STD was slightly higher in the non-injured group. CV was in general higher in the non-injured group and tended to increase in this group, whereas it decreased in the counterpart. The exponent α decreased in both groups from the beginning to the middle, and then did not change for the non-injured, but increased for the injured runners. This exponent was higher in the non-injured group. They started their run with a value of 1.19 and leveled down to 0.96. The injured runners displayed 0.92, 0.68, and finally 0.77. The high values in the non-injured group might be due to the fact that these were recreational runners. Their study also suggested an increase in variability and a reduction in the exponent α during fatigue. Injury may narrow the degree of freedom and, therefore, limit the range of motion and possible adaptations. A smaller amount of variability with less predictability seems to be preferable over all, but high qualities in running form and injury can cause similar behavior (seen in measures of variability).

2.2.3 Summary

With high caution to the constrains of *DFA*, it appears as a promising method revealing non-linear dynamic strategies of the motorsystem, or at least the mechanical system. Furthermore, the amount and the structure of the variability have to be considered. Most studies were conducted analyzing the human gait, for running treadmills have been used. In over-ground running, the stride interval has been of interest. Information for stride length are not present.

2.3 Measurements in outdoor scenarios

As mentioned in the introduction, since the beginning of "Ubiquitous Computing", there is a rising wealth in the technologies supporting athletes during their training. The focus of this section is directed towards technologies and methods for the estimation of stride parameters in outdoor running. A closer look will be spent on (mobile) systems, which incorporate Global Position System (GPS) and/or inertial sensors such as accelerometers and gyroscopes. Each of these technologies has its merits and drawbacks. Systems deploying a combination of them known as inertial measurement units (IMU) allow for a complete estimate of the kinematics i.e., its linear and angular components, and improve its accuracy and completeness.

2.4 Sensors

There are several Global Navigation Satellite System (GNNS), i.e. the Global Navigation Satellite System (GLONASS) of the Russian federation, Galileo of the European Union, Compass of the Republic of China, and the Global Position System (GPS) of the USA. GPS is part of the US-American Navigation System for Timing and Ranging

(NAVSTAR) and its official name is NAVSTAR GPS. The satellites and the tracking stations on earth are synchronized with an atomic clock. The satellites continuously send a data stream to the earth. A GPS receiver determines the distance between itself and a satellite by evaluating the time of arrival and send i.e., the distance equals the product of the signal traveling time and the speed of light (Larsson, 2003). At least four satellites should be available for each position of the world, in order to determine the 3-d position. In order to achieve centimeter accuracy for determining the position in walking and running, a high-end GPS with $5\,Hz$ to $20\,Hz$ is required (Terrier & Schutz, 2005). High-end GPS makes use of another tracking station with a known GPS-position, which allows for the estimation of the position error. For example, Terrier *et al.* (2005) measured the stride parameters such as the stride speed, the stride length (SL) and the stride rate (SR). With the help of such a high-resolution GPS reciever (attached to the head, and a sample rate of $10\,Hz$), they could segment the position data into single step by analyzing the vertical position. Low-grade receivers as build in smart-phones do not provide this accuracy and precision. The segmentation has to be organized towards other sensors. Furthermore, low-cost GPS receivers have an accuracy between $2\,m$ and $15\,m$ depending on the available satellites. Adverse weather condition (e.g., clouds and fog) and terrain (e.g., wood, high buildings, etc.) reduce the precision. A pure GPS approach to the estimation of stride parameters and its variability would probably fail.

Direct measurements of human movement by accelerometry allows for studies or applications concerning kinematics or associated analyses outside the laboratory. Thereby, its low power consumption, small size and low costs (Godfrey *et al.*, 2008) serve a useful purpose. An accelerometer sensor can be imagined as a cubic box, in which is a ball. If the box is moved, the ball is pushed against the wall opposing the direction of movement. The ball hits the wall with a force according to Newton's 2^{nd} law of motion ($F = m \cdot a$). This model is extended with a spring linking the wall and the ball. This spring opposes a force according to Hook's law (force equals the product of spring deviation x and the spring constant). The measured acceleration equals the force of the spring divided by the mass of the ball ($a = k \cdot x/m$) (Kavanagh & Menz, 2008; Mathie *et al.*, 2004). Accelerometers are available as uni-, bi- or tri-axial sensors and can be grouped into fluid, reluctive, servo, magnetic classes, and more commonly in the analysis of human motion, in classes of strain gauges, piezoelectric, piezoresistive and differentiable capacitor accelerometers (Kavanagh & Menz, 2008; Yang & Hsu, 2010). Acceleration signals are superimposed by vibrations. There though is noise from the hardware and, moreover, when analyzing humans, soft tissue causes additional vibrations. The presence of the gravitation acceleration (g) and its projection on the respective axes has to be taken into account for evaluation of the acceleration data. For example, an uni-axial accelerometer can provide the estimation of the pitch, the angle between the horizontal planes of the earth and sensor. Let‘s assume that the respective axis is in direction to earth (acceleration a_y). The angle θ between the earth's gravity and the rotation around its horizontal axis can be calculated by $\theta = \arccos a_y/g$. The pitch is then $theta + 90°$. Bi- and tri-axial can provide further estimation of roll and yaw by similar trigonometric functions. One solution is to calculate the magnitude of the three accelerations euclidean distance. The arc cosine function with the acceleration of the corresponding angle of interest divided by the magnitude yields the angle of interest. A simple method to account for the earth gravity in a tri-axial accelerometer is the subtraction of it from the magnitude. However, in reality noise, gravity, other forces, and insufficient sensitivity of the sensor disturb the estimation of the orientation, and as well, the estimation of the velocity and position by integration methods. Methods are required to handle these factors. For example, there are several solution to remove earth gravity, and thus to consider the accelerations in the inertia frame of the sensor. For this purpose, stochastic filters and wavelet decomposition have been applied (Han *et al.*, 2009; Aminian *et al.*, 2001). Low pass filters with cut-off frequencies ranging from $0.2\,Hz$ to $0.5\,Hz$ have been shown to be suitable in the analysis of walking gait (Han *et al.*, 2009). The use of special calibration techniques have been suggested especially in the clinical environment by e.g., Lai *et al.* (2004), Morris & Paradiso (2002), Lotters *et al.* (1998) and Ferraris *et al.* (1995).

In walking gait analysis, accelerometry has been successfully applied to detect features in the stride kinematics. For example, Jasiewicz *et al.* (2006) placed an accelerometer at the foot, whereas Saremi *et al.* (2006) confirmed thigh and the foot as suitable body locations. Another example placing the sensor at the trunk is the study of Zijlstra & Hof (2003). They determined the step length with an empirical formula, see eq. 2.21, for each step.

$$steplength = 2 \cdot \sqrt{2lh - 2h * 2} \tag{2.21}$$

$h \ldots$ vertical difference of COM, $\quad l \ldots$ leg length

Other body locations to place the sensor(s) were the head, upper trunk, and mandible. In over-ground running, Neville *et al.* (2011) determined SR with tri-axial accelerometer with a sample rate of $100\,Hz$ attached to the back of the runners. After removing the baseline of the data and applying several filters, they counted the zero crossings of the signal in the vertical axis, i.e. steps with time a time line yielding SR. SR and the real speed had a strong correlation. Lee *et al.* (2010b) placed a tri-axial accelerometer with a sample rate of $100\,Hz$ at the sacrum of ten national runners. From treadmill running, they determined the motion of the center of mass during the run. They emphasized that asymmetry between left and right body side "may arise with the development of muscle fatigue and/or changes in exercise intensity" (p.559). They could confirm the expectations from other studies that the oscillation of the center of mass decreased as speed increased. Their system was able to "detect small changes in gait symmetry" (p.569). They developed this approach further (Lee *et al.*, 2010a) and determined the stride time, step time and stance phases. The foot strike was detected in the peak antero-posterior acceleration; a smaller peak of this acceleration indicated the toe-off. Left and right foot could be distinguished with the analysis of the mediolateral accelerations. Differences to the reference system were negligible small (between $-0.024\,s$ and $-0.023\,s$). Auvinet *et al.* (2002) used a tri-axial accelerometer placed at the trunk, in order to detect the initial contact, mid-stance and the toe-off event in over-ground running at middle-distance speeds (approx. $5.16\,m\,s^{-1}$). The same intention had Bergamini *et al.* (2012), but failed for the estimation of temporal parameters (based on the stride events with the acceleration and the first and the second derivatives) during sprint running. They speculated that the position of the accelerometer might be one reason therefore, but more probably seemed to them, the high speed compared to studies in walking or with lower speeds. Taken collectively, accelerometer based analysis systems are able to detect features in walking and, therefore, are present in the clinical environment, however, the estimation of stride kinematics in running is challenging due to higher speeds.

Gyroscopes can be considered as the complement of accelerometers by providing the orientation. There are mechanical and optical ones. The traditional gyroscope is a wheel acting as a rotor, which can spin in all dimensions. The spin axle of the wheel is inside a ring. To this ring, a gimbal, an outer ring, is attached in the plane of the wheel. The frame of the gyroscope fixes the gimbal, in a way that all axes cross the middle of the wheel aligning all axes in 90° to each other. When the frame of the gyroscope is fixed to a ground and the rotor is spinning, then the gyroscope shows precession and nutation. If the frame is free and the gyroscope moves in all three axis, then the spin axle of the wheel will force to maintain in its orientation. These effects are used to determine the angular velocity and the angle. In contrast to accelerometers, gyroscopes are more precise while responding faster to changes in the angle; they are not sensitive to vibrations, but suffer from a drift in the angle, which develops quickly over time. This drift can be compensated e.g., with the help of a accelerometer and filters, see next section. Gyroscope are also available as uni-, bi- or tri-axial sensors and all have been deployed in the analysis of walking gait analysis by often placing them to the shank e.g., in order to detect temporal gait parameters through events such as initial contact, stance and/or toe-off (McGrath *et al.*, 2012; Greene *et al.*, 2010a,b; Tong & Granat, 1999; Jasiewicz *et al.*, 2006). Therefore, gyroscope data has to be firstly calibrated to the conditions at the beginning. A (low pass) filter is then applied on the angular velocity. The gait events follow a pattern and also with threshold values they can be detected. Another purpose is the estimation of angular velocity and/or the angles of the foot during the gait by placing the gyroscope at the (rear or hind) foot as e.g., Pappas *et al.* (2004) and Brauner (2010) did. Miyazaki (1997) placed a gyroscope near the knee at the thigh, then determined the angle of the leg perpendicular to the ground. This and the leg length was comprised in a simple pendulum model that estimated then the step length. Bergamini *et al.* (2012) demonstrated that sprint running challenges the feature detection for gyroscope based assessments, too. A wavelet based approach smoothing the angular velocity allowed for the segmentation of the gait cycle. Though gyroscopes have been shown to serve as a worthy tool in the analysis of gait while featuring mainly temporal parameters and angle velocity derivatives, but likewise with the accelerometers higher speeds compound the detection of gait events.

2.5 Sensor fusion

In order to cope the drawbacks of the single sensors, they have to be deployed in a measurement unit together. For example, to describe the motion of such an unit in the three-dimensional space, six degrees of freedom (6-DoF) i.e., three translations and three rotations, are necessary. The translations is usually measured by three orthogonal aligned accelerometers, whereas the rotations are measured by three orthogonal aligned gyroscopes. These inertial sensors, however, can only give relative information about their position and orientation in the navigation frame. The (non-linear) error of the determination of the absolute position and orientation depends from (hardware and environmental) noise, sample rate and the integration methods (due to gravity, initial values, bias, discrete signal). That is why, information from the environment (navigation frame) has to be involved, too. A common 3-d IMU comprises also a magnet sense, which provides the azimuth. Algorithms, which fuse the data from disparate sources (sensors, constraints), are strategies to handle these probably inconsistent data, in order to determine the kinematics. There are a variety of strategies for different purposes, for example, the Pedestrian Dead Reckoning (*PDR*) approaches, which attempt to estimate the position over time of pedestrians (or any objects) with the help of different sensors and constraints. The most common fusion algorithms is the Kalman filter, which makes use of a stochastic model of the system and the environment. This filter estimates the state of the system by noisy measurements in two steps. First, the filter predicts the new state of the system (e.g., the motion of the pedestrian) with respect to the certainty (i.e. the variances and means of the measurements) of the motion. Second, the filter updates the stochastic variables (i.e., the variances and the means). In the following, several approaches combining inertial sensors in the gait analysis will be introduced.

There are approaches as well that combine data of accelerometers and gyroscopes to improve the detection of gait events and determine further kinematics of the lower extremity. For example, Takeda *et al.* (2009) and Liu *et al.* (2009) placed several IMUs at the lower extremity and determined e.g., the knee and the foot angles. Mariani *et al.* (2012) used the angular velocity to segment the cycle of walking gait and determined further features such as the stance phase with the help of acceleration based data.

Yang *et al.* (2011) used an IMU placed on the shank to determine the running speed in treadmill running. They stated that the integration error due to the sensor bias increases with speed. Therefore, it is important to segment the gait cycle and proceed the integration only in short periods. They found that the segmentation was different than in walking. Their method estimated the running speed with a root-mean-square error of 4.1%. In walking analysis, Mariani *et al.* (2010) estimated among the temporal stride parameters SL with an IMU placed at the rear foot. First, the stance phase was detected with the angular velocity near at zero and and its end referred to the initial phase of each cycle. During the stance the foot was considered being motionless, see next paragraph for details. Second, the gravity in acceleration data was canceled. Third, the drift of the (trapezoidal) integration was removed "by subtracting a sigmoid-like curve modeled based on a [monotone piecewise bicubic (p-chip)] interpolation function" (p.3002). Finally, SL equaled the distance between two consecutive stance phases. However, in the majority the Kalman filter has been applied to determine SL. The difference to the aforementioned methods is the integration of additional information from the environment, which is used for the correction of the heading i.e., the direction of the current step. As representatives for this approach to determining stride parameters, four studies will be introduced. Previously mentioned and other studies made use of a simple mechanical model of the walking gait or estimated the spatial parameter with linear regressive models. Conversely, Koese *et al.* (2012) mounted an IMU featuring a tria-axial accelerometer and two bi-axial gyroscopes to the right side of the pelvis and applied a Kalman filter based on both accelerations and angular velocities to obtain the right and left step length from double integrating the acceleration data. It was assumed that the step length equals the traveled distance of the pelvis. The gait events such as heel strike were detected in the wavelet decomposed acceleration data by using to thresholds being a proportion of the magnitude. The integration was performed on each cycle with caution to the pelvis rotation at the end of an gait cycle. The average error ranges from $-2.6\,\%$ to $2.9\,\%$ and from $0.4\,\%$ to $2.6\,\%$ for right and left step length, respectively. Kim *et al.* (2004) proposed a method, which assumed a linear relationship between SL, SR and the vertical acceleration. To better detect the initial contacts they applied a pattern for vertical acceleration data. Therefore, they did not rely on constant threshold values, but could dynamically (for each step) adjust them to the speed and to the data of the uni-axial, antero-posterior accelerometer. The walking direction was estimated with uni-axial gyroscope and a magnetic compass attached to the foot. Gyroscope and magnetic compass were compensated by the Kalman filter. SL was determined by the experimental eq. 2.22. The magnetic compass is susceptible to "unpredictable external disturbances" (p.277). The error was 1 %, 5 %, and 5 % for the step detection, traveled distance and heading error, respectively.

$$SL = 0.98 \cdot \sqrt[3]{\frac{\sum_{i=1}^{N} |a_i|}{n}} \tag{2.22}$$

n . . . number of strides, a . . . acceleration

When fusing inertial sensors, criteria have to be considered that help to lower the drift of the integration of the accelerations. Without corrections the error of the velocity would increase linearly with time while the error of the position increases quadratically (Fischer *et al.*, 2012). All studies so far exploited the cyclic nature of the gait to perform the integration for each step separately. If a Kalman filter is applied, then the error has to be estimated. This error also helps to correct the calculation of the desired parameter. In walking gait analysis, it is assumed that the velocities are zero at the mid-stance. This has become famous as the zero update velocity assumption (ZUPT). For example, the evaluation of the angular velocities and the accelerations indicate a stance-phase, however, the sensor values are not zero. The error then refers to the current sensor values. Two reasons cause this error. First, the sensor sensitivity and the noise from the strapping to the shoe or the skin. Second, the foot's velocity is not really zero, because it moves partly. In walking, however, Peruzzi *et al.* (2011) proved that the hind and rear foot cause minimal error, but this increases with walking speed. Positioning the IMU at the lateral aspect of the rearfoot or the calcaneus "showed a minimum velocity and limited dependency on gait speed and limited timing variability" (p.1993). With eq. 2.23, SL can be determined for noise-free antero-posterior acceleration data for each step. The last example on *PDR* is the method of Fischer *et al.* (2011), which comprised only one IMU mounted on the hindfoot. After having detected the stance phase, their algorithm estimates the position and velocity error and then calculates the error co-variances. Thereafter, the velocity and the position is corrected. They made use of a tria-axial accelerometer and gyroscope and demonstrated that this method was stable for walking.

$$SL = \int_{t=0}^{T} \int_{\tau=0}^{t} a(\tau)d(\tau)dt + v_0 T \qquad v_0 \ldots \text{antero-posterior velocity at beginning} \tag{2.23}$$

$$a \ldots \text{magnitude of acceleration} \quad d \ldots \text{delta} = 0.01 \text{ (time step of integration)}$$
$$T \ldots \text{duration of integration} \quad t, \tau \ldots\ldots \text{current time step during integration}$$

2.6 Summary

A variety of approaches comprising accelerometers, gyroscopes, magnetic compass and GPS have been undertaken in gait analysis. It appears that the determination of temporal stride parameters is a less changeling than the spatial ones. Especially, for over-ground running, there is no system present that can determine the spatio-temporal stride parameters with sufficient accuracy, in order to estimate stability and variability.

3

Outline and thesis

MC system for runners

From the literature, the author lent credence that evaluating the stride kinematics of runners may support the assessment of their performance and injury risk. Thanks to Pervasive Computing the technological requisites offer possibilities to implement an MC system for runners. Therefore, the first purpose of the thesis was an implementation of an MC system for runners. Comprising low-cost and hence low-resolution sensors, this system should allow for the determination of basic kinematic parameters displaying the speed, the stride rate (SR) and covered distance. Moreover, the feasibility of the MC system should be evaluated with respect to the accuracy and precision of the parameters and the possibilities to draw conclusion upon the stride kinematics and its feedback provisions.

IMU/GPS based stride parameter determination

While temporal parameters in the stride kinematics have been studied, little is known about spatial parameters in over-ground running. The stride length (SL) appeared as a crucial indicator of performance. Therefore, the second purpose of the thesis was to extend and evaluate a PDR method for the suitability in running, in order to instigate further investigations of stride kinematics in over-ground running. It is sought to achieve a suitable accuracy and precision to facilitate the analysis of variability and stability in the running cycle.

Behavior of stride kinematics in a 5 km time-trail

Stride kinematics have been found of importance to characterize athletes during training. However, the development over the time course has not been substantially declared in outdoor running due to methodological limitations capturing few strides. The thesis aimed to shed light on this with the help of a measurement system gathering approx. 20 % of all strides. Middle-distance running is practical for recreational runners in terms of time exposure and physical strain, and thus a typical use-case for the MC system. Therefore, the predictions of the model according to Saziorski *et al.* (1987) for elite runners were compared to the behavior of stride parameters of recreational runners in a 5 km time-trial. Furthermore, the feasibility of this model within an MC system was examined. Regarding performance, it was of interest to identify the relationship between SL and SR in a full-exhaustive run. The following hypotheses (H) were examined:
$H1$: The influence of SL on performance is greater than that of SR.
$H2$: SL decreases with fatigue.
$H3$: SR compensate in the second third of the time-trail.
$H4$: Both, SL and SR decrease in the last third.

Little information is available about the variability in over-ground running. In elite runners, the coefficient of variation (CV) of the running speed is approx. 3 %. As long as fatigue does not dominate, the run is optimized towards performance while reducing variability according to the minimum energy hypothesis. With the onset of fatigue, the brake-mechanism fades in according to the variability-overuse injury hypothesis leading to an increase of variability.
$H5$: CV of the stride parameters in recreational runners is greater than 3 %.
$H6$: Variability expressed by the standard deviation (STD) increases during the race.
$H7$: Stride pattern changes with fatigue (e.g. shorter SL and higher SR). Nevertheless, both SL as a function of stride speed (SS) and SR as function of SS display a monotonic behavior.

Indicators of fatigue

The influence of the speed distribution over the race has little meaning:
$H8$: The parameters SLn and SRn (SL and SR normalized for SS) display the same overall behavior as SL and SR.
$H9$: The SR index (SLI) and the SR index (SRI) demonstrate that recreational runner make use of the SL to induce changes in SS.
$H10$: In contrast to walking, the running relation (SL/SR) is not constant.

Heart rate (HR) is assumed to level out above anaerobic threshold ($T2$):
$H11$: $HR : 1/CT$ index increases with fatigue.

Leg stiffness decreases, the rate of force generation for propulsion decreases and hence the contact time (CT) increases. Overall SR and ST were expected to slightly decrease and increase, respectively. The influence of running speed diminishes as fatigue sets on.
$H12$: CT increases.
$H13$: FT decreases.
$H14$: The duty factor ($CT \cdot SR$) increases, too.
$H15$: The relation FT/ST decreases.
$H16$: CT normalized for SS (CTn) increases.
$H17$: FT normalized for SS (FTn) decreases.

Over-striding was mentioned as an inappropriate running technique mainly in inexperienced runners with regard to a lowered efficiency and higher risk of injury:
$H18$: Recreational runners tend to over-stride i.e., $SL > SL_{optimum}$.
$H19$: Recreational runners tend to lower than optimal SR i.e, $SR < 1.5\,Hz$.

The risk of injury is associated with a decreasing leg stiffness and higher (vertical) accelerations:
$H20$: Vertical accelerations increase during the race.

With the onset of fatigue, the correlation between speed and accelerations weakens:
$H21$: For SS normalized vertical accelerations increase during the race.
$H22$: For the beginning normalized vertical accelerations increase during the race.
($H20$-$H22$ have not been examined within the thesis.)

Stability of stride parameters in a 5 km run

For walking and treadmill running, it has been demonstrated that ST contains fractal-like behavior, hence long-range correlations do persist. The stability was degraded at speeds other than the preferred ones. There was no study demonstrating this behavior for SL in over ground running. At all, stability of the running cycle in all stride parameters is assumed to decrease with fatigue.
$H23$: The fluctuation of the stride parameters exhibits a long-range, fractal-like correlation pattern under fatiguing running conditions.
$H24$: When individuals fatigue near to full-exhaustion after the middle of the run, the fluctuation pattern becomes random.

The ultimate goal was to relate the behavior of the stride parameters, the amount and the structure of variability to each other, and though to set up markers of fatigue to assess performance and the injury risk.

4

A Mobile Coaching system for runners

4.1 Introduction

Physical activity or even exercise has been appreciated contributing to a healthy life style. The assistance of technical devices has been registered in health and medical services, leisure activities and in sports—as early humans extended their room for maneuver with tools. The purpose of this study was the development of a technical system, which would support runners in their training. Therefore, a Mobile Coaching (MC) system for runners was developed. Experts should have the ability to observe training sessions live on the Internet and assist the runner through feedback messages, even if they are remote. Runners should be equipped with hardware, which does not disturb their training. In this meaning, the equipment had to be small, lightweight and easily attachable. The energy consumption had to be guaranteed at least for a typical training session in middle-distance running. The server-sided routines should also be prepared for feedback generation.

4.2 Architecture

The MC system for runners is built on an interactive bidirectional communication technology. As long as an Internet connection is available, both the experts and the athlete are able to communicate with each other. Sensors attached to the runner collect data, which are transferred via a body sensor network (BSN) to the runner's mobile

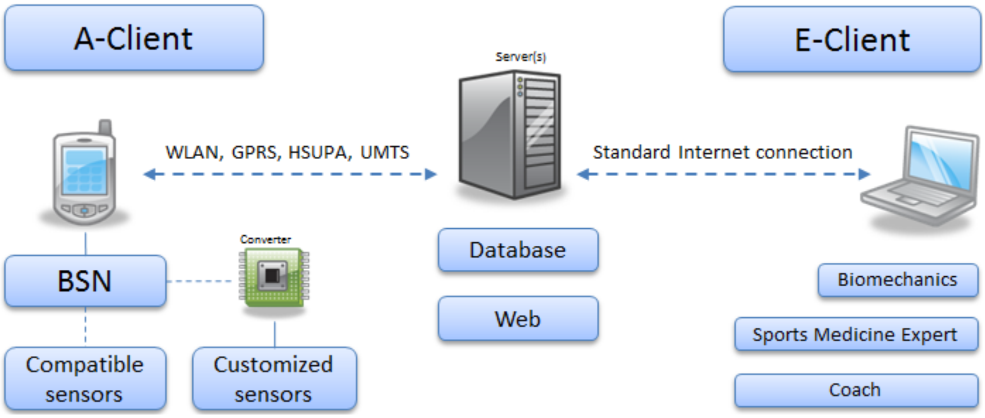


Figure 4.1: Architecture of the MC system for running

Internet device, the A(Athlete)-Client. The data are pre-processed and forwarded to a database server via the Internet. A Web interfaces, the E(E-Client), is offered to the experts. This interface allows to observe the current training, review previous sessions or send feedback messages to the athlete(s). A host server has access on the training database and hold software components, in order to analyze the data and establish the communication between the A- and E-Client. A BSN was chosen that can integrate a plenty of sport specific sensors and has a lower energy consumption compared to the Bluetooth protocol. These sport specific sensors offer an interface to the BSN. In order to further integrate various analog sensors, the micro controller (NEON, Spantec GmbH, Linz, Austria) is foreseen to convert the analog into digital data according to the protocol of the BSN. Kinematic data is obtained from the low-grade GPS-receiver of the smart-phone and also from a customary sensor obtaining the stride rate, called footpod. The heart rate sensor provides the rate of the heartbeats. In general, the MC prototype can be extended to any sports application with sensors which are supported by the BSN, or which are customized for the micro controller, see fig. 4.1.

4.3 Implementation

The implementation used in all experiments of the thesis comprised the following components: a host computer providing a web and a database server, the MC applications on the host server, a smart-phone with an MC application, sensors cable-freely connected to a smart-phone, and a special adapter managing wireless networks. The applications on the host computer run on an Apache®web server (Apache Software Foundation, http://www.apache.org) in a traditional combination with PHP 5 (The PHP Group, http://www.php.net) and a MySQL®database (Oracle Corporation, http://www.mysql.com). The database schema offers the management of users, sensors and training sessions and plans. Sensors are stored with their properties (e.g., network key, channel id, transmission and device type), in order to facilitate the configuration of the A-Client. A user can represent an athlete or an expert. Training plans consist of training sessions and the latter of exercises. A minimal configuration for a training though consist of one exercise; otherwise a training was a sequence of exercises. For example, an exercise can be a warming-up, a sprint, any interval training, or a combination of these exercises. Runners are instructed how to perform an exercise with a suitable name and a description of the exercise. Each exercise is also associated with sensors, which then are used during this exercise. It is possible to fill in values for these exercises-sensor combinations, in order to e.g., feed server- or client-sided routines targeting the runners in appropriate training zones. For example, for moderate training via the heart rate 60% to 70% of the maximal heart rate was assigned for this exercise. The front end is a web application, which experts can access with an Internet browser via the Hyper Text Transfer Protocol (HTTP). First, this web application is used to manage the users, sensors, training sessions and plans; second, it offers to observe and analyze the training data; and third, feedback can be sent to the A-Client. An application on the smart-phone manages the training sessions. It also communicates with the back end to initiate the data exchange. The back end of this MC implementation handles the communication between the A-Client and the database. Furthermore, it offers all the functions to manage the training sessions e.g., start and stop the training, transfer data of the user and sensors, see fig. 4.2.

A specific BNS was put to use (ANT™, Dynastream Innovations, Alberta, Canada, http://www.thisisant.com). Especially, its extension ANT+ was promising in offering sport specific sensors by becoming a standard protocol i.e., sensors can be easily integrated and combined. There was no smart-phone available supporting the ANT protocol. Several solutions were tested to establish compatibility. For example, for a Windows Mobile 6.5 based smart-phone (HTC HD, HTC Company, Taoyuan, Tai-

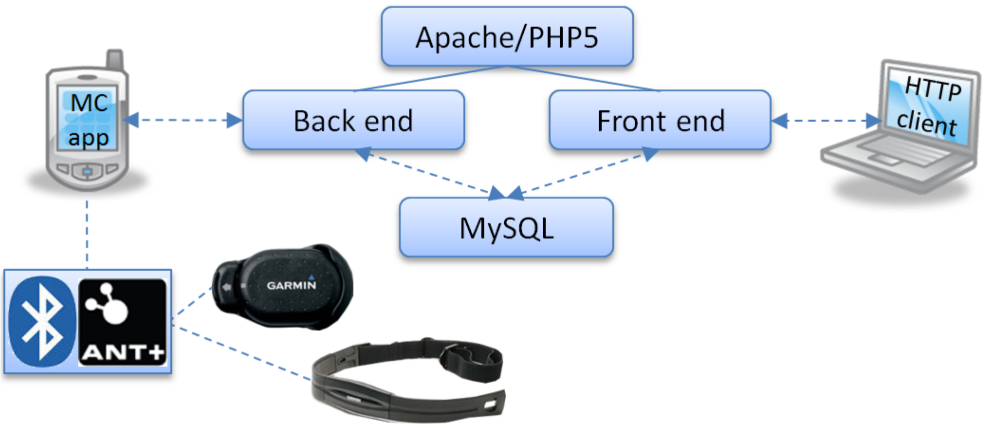


Figure 4.2: Implementation of the MC system for running

wan) a micro SDIO-Card were modified by comprising an ANT-chip. Therefore, the application on the smart-phone (programmed in C#, Microsoft®.Net) can set-up an ANT-network. ANT-based networks have a range up to 15 m. This configuration reduced the range of the network dramatically, probably due to the surrounding metal parts of the smart-phone, which disturbed the magnetic field of the ANT-chip. A temporary version of the client (used in the experiments of the thesis) makes use of an external ANT-chip. This is connected to a common smart-phone (Samsung®Galaxy 3, GT-I5800, Android®OS 2.2) via the Bluetooth®protocol (http://www.bluetooth.org). Therefore, a Bluetooth-to-ANT adapter was individually manufactured incorporating an ANT (Nordic nRF24AP2) and a Bluetooth®(OBS411) module, see fig. 4.3 and 4.2. This configuration ensured a battery life of approx. one hour and worked stably.

The MC system for runners comprises a GPS receiver to determine the position of the runners, an uni-axial accelerometer, the footpod, to determine the speed, the number of strides and the stride rate, and the heart rate to estimate the intensity of the training. In order to perform a training session, it has to be created via the front end in before, the heart rate belt (HRM1, Garmin Ltd., USA) has to be worn on the rib cage and the footpod (SDM4, Garmin Ltd., USA) has to be attached on the hind foot. During the training the smart-phone has to be attached on the upper arm with an armband. Starting the MC application on the smart-phone the athlete is asked to log in, in order to load one's user profile and the current training data. According to the

Figure 4.3: Bluetooth-to-ANT adapter

selected exercise, the referring sensors were searched and once successfully registered the start button will be unlocked. Training data is recorded as the user pushes the start button. The graphical user interface on the smart-phone provides the athletes with the current values of the heart rate, distance and current speed. Feedback routines can evoke vibration, speech, music and display light. Training data is collected at the smart-phone, buffered and sent in an interval of $10\,s$ to the database. The buffer (a local linear growing file) is necessary when the Internet connection breaks up. Heart rate measurements were able to determine beat-to-beat measurements, which allowed for the analysis of the heart rate variability. The heart rate sensor sends a data package each second to the client. The GPS receiver operates on a sample rate of $1\,Hz$, while the footpod operates on a sample rate of $4\,Hz$.

The traditional components used for this approach endows the MC system with stability. However, the realization of the feedback generation has been challenging because instant feedback was claimed without an observer or event driven pattern of the A-Client application. Both of these patterns would have caused further load on the smart-phone and data traffic. The relinquishment could be compensated with a compromise i.e., feedback messages are read from the database table after data has been sent into database. As aforementioned, the back end collects the arriving data from the A-Client and forwards them into the database. Experts using the web client can access these data through the front-end. For example, the experts can inspect the position of the runner on a map, the current speed, distance and heart rate. Tables, basic statistics and charts (programmed in JQuery and OpenFlash) are presented to the experts as analysis tools. Feedback is posted while inserting it into the feedback table, which functioned as a stack. After inserting sensor data, the last feedback message will be taken from table and then this message will be marked as read after being fetched by the A-Client, see fig. 4.4.

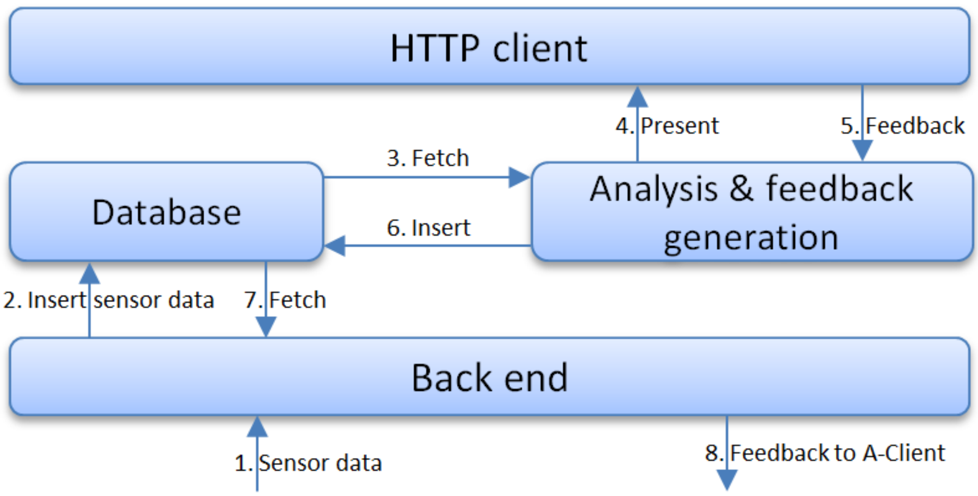
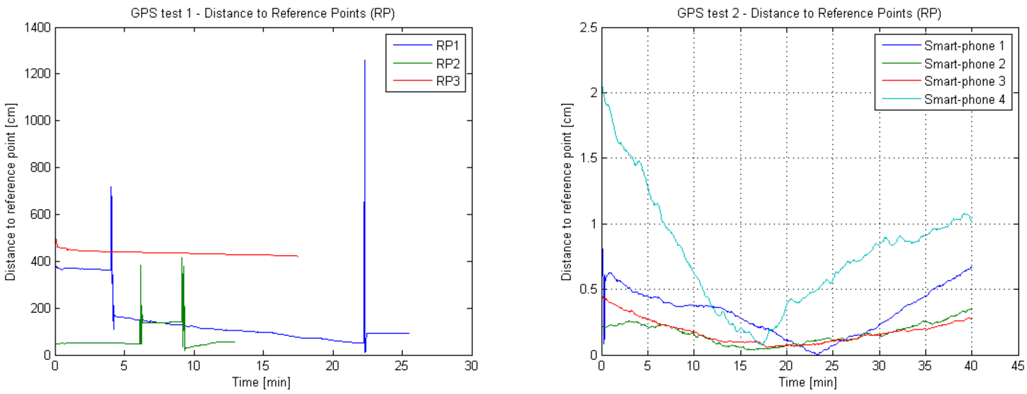


Figure 4.4: Analysis and feedback generation of the MC system for running

4.4 Tests and Results

Outdoor and indoor tests confirmed the expectation of varying data transfer and hence the need of a buffer. However, it could be proven that the MC system can transfer up to about 1450 values per data packet per 10 s stably (8700 vales per 60 s). The linear growing buffer is a simple and effective method until this limit is not overreached. Unfortunately, no analog sensors with a higher sample rate could be tested, because the analog/digital converter was not ready for operation. The MC system for runners requires about 60 and 360 values per 10 s and 60 s, respectively—far below this limit. In order to prove the MC system for its data transmission, reliability of the kinematics and usability in running, further tests were performed.

The heart rate measurements are based on beat-to-beat measurements pertaining to the sensor specification, which were not further proven. Heart rate and the intervals between the beats can be gathered, and therefore the heart rate variability determined. In contrast, there was no knowledge about the reliability of the kinematics determined by this configuration of the MC system for runners. GPS was known to fluctuate with dependency from several conditions; the footpod was expected to suitably measure the stride derived parameters over several strides. This sensor could be calibrated to a runner's typical stride length (SL); this combined with the stride cadence allowed for



(a) (b)

Figure 4.5: (a) A smart-phone was subsequently placed on three different reference points (b) Four smart-phones were placed on reference points at the corners of the field inside an athletic track.

the determination of speed and distance—as long as the runner could maintain SL. The documentation of the footpod ensures a reliable measurement at SR of approx. $2\,Hz$. Pilot tests proved this true for average values over a series of strides. Nevertheless, it was not known whether or not a stride-to-stride based analysis can be conducted. Therefore, the GPS, the footpod and the combined derived parameters were examined.

In order to compensate the fluctuations of the GPS, it was aimed to establish a reference station for improving the position by an error determination. Two tests were conducted to examine the behavior of the GPS signal of fixed GPS receivers on reference points. For test one, a smart-phone was placed at three different locations sequentially for about $20\,min$. The locations were on the athletic track, which was used for all experiments. For test two, four smart-phones were placed at the corners of the field inside the athletic track for about $40\,min$. Finally, several determinations of stride parameters were tested and compared to the results from test three. This test was a video analysis of three subjects running at three speeds each two laps. These tests were part of a pilot study (Kremser, 2011).

From fig.4.5 it becomes apparent that the raw GPS can have large outliers, perhaps due to the replacement of satellites while calculating the position. In the beginning and after each displacement of the smart-phone, the calculated position tended to the real position over the time but never reached. The overall error was $2.18\,m \pm 50.52\,m$ when the outliers were not removed, see fig. 4.5 a. From test two, it could be seen that distances to reference points of the four smart-phone differed considerably. The determined distances to the reference points of the four smart-phones varied in a similar manner i.e., from the beginning to the middle of the test, there is a downward trend, and then after the middle, an upward trend. There were no outliers. The overall error was $0.36\,m \pm 0.20$. Both tests illustrate the variability of the measurements.

When the GPS and the footpod were fused to determine the SL, two factors influenced the calculation adversely. First, the fluctuations could not be controlled by the reference station. Second, the footpod worked stable but missed several strides. Within $400\,m$ 52 strides were miscounted when comparing the extrapolation of the video analysis ($30\,m$ of the long side of the athletic track) and the counts of the footpod. Moreover, it remained unknown when these failures occurred and therefore, could not be taken into account. With some exceptions, the results for SL determined by the three methods were similar. For example, ignoring the outliers would result in a difference between video analysis and the fused GPS and footpod determination of $0.26\,m \pm 0.12\,m$, with outliers it was $50\,m \pm 69\,m$, see fig. 4.6.

4.5 Discussion and conclusion

Training relies on the principles of learning. A training can evoke a learning process, which is responsible for the desired improvements. In order to develop qualities in a movement task proprioceptive and exteroceptive feedback mechanisms have to be employed and optimized. Although cognitive efforts are made, a majority of the adaptations in cognitive or motor behavior is achieved by unconsciousness processes, and thus is not directly accessible to the learner. If feedback from outside is given in a suitable way, then it has been shown to improve learning, hence self-awareness and performance. Technical equipment might make some aspects of the learning process available to the coach and athlete, and likewise to people in other fields. For exam-

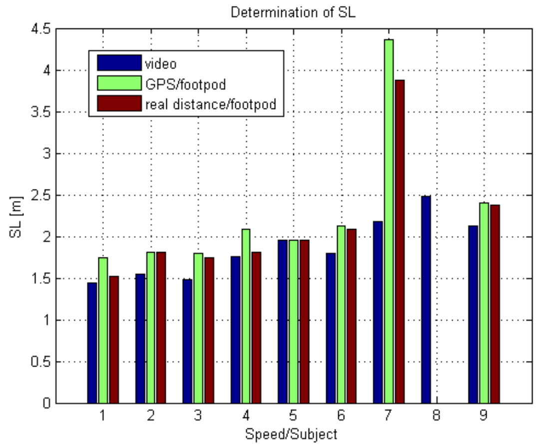


Figure 4.6: Determination of the SL by a video analysis, fusing the GPS and the footpod and the distance of the athletic tack divided the counted stride by the footpod. Three subjects ran at three speeds ($2.0\,m^{-s}$, $2.5\,m^{-s}$ and $3.5\,m^{-s}$) and two laps of $400\,m$ athletic track. Three groups of bars represent one speed. Within the groups the sequence of the bars refers to the subject.

ple, a result of a performance might be more efficiently appreciated when the inner performance i.e., the effort can be related to it. This MC system allows for the presentation of a variety of among others physiological and biomechanical data. For example, the heart rate variability can be evaluated with existing methods. Nevertheless, this generic approach can be adapted to a specific system and use-cases. The MC system for runners provides runners and experts with the standard parameters of today's training and further includes the ability to feedback messages and the analysis of the training.

In future, it is foreseeable that more sensors with higher sample rate might be incorporated. The buffer of the A-Client might be then replaced more sophisticated techniques such as a local database, which can synchronize data after a data stream interruption. In one of the test implementations we could make use of a local database. The drawback was the higher load on the A-Client. Smart-phones of the next generation are expected to be more powerful and will handle this issue. Also then it is considerable to select features at the A-Client and only send them to the database. As expected, GPS and footpod derived parameters support the training of runners by providing average values over several strides. No stride-to-stride based analysis seems to be realizable with the current configuration. Therefore, a further approach was undertaken based on sensors with higher sensitivity and more dimensions, see next chapter.

5

IMU/GPS-based stride-parameter determination

5.1 Introduction

Analysis of human motion by inertial measurement units (IMUs) and GPS has become attractive since sensors can be placed in sports equipment nearly without disturbance (Baca *et al.*, 2009; Godfrey *et al.*, 2008). Non-differential GPS-measurement is a common tool to log distance and speed of running using mobile devices. Emphasis lies in the motivational character e.g., publishing activity logs in social networks (Zong, 2008). The present study goes further and concentrates on stride parameters in a biomechanical context. These parameters are usually bound to laboratory environments and usually gathered by performing on treadmills. Other solutions are found in using e.g., D-GPS (Terrier *et al.*, 2000). Tan *et al.* (2008) proved that the integration of GPS and IMU can determine average speed and intra-stride variation in stride rate and vertical displacement. Furthermore, the movement analysis by accelerometers and/or gyroscopes promises to be successful in establishing stride-to-stride measurements in running. For example, Neville *et al.* (2011) reports that stride rate and speed determined by accelerometers can be used for monitoring track and field athletes. Researchers like e.g., Lee *et al.* (2010b), Auvinet *et al.* (2002) and Bergamini *et al.* (2012) detected the initial contact, the contact time and the stride time of consecutive strides with the help of accelerometers. For the same purpose, gyroscopes were employed by others e.g., McGrath *et al.* (2012), Greene *et al.* (2010a) or Brauner (2010). However, these analysis systems are designed for the use in walking or on treadmills. To date, movement stride based analysis in over-ground running is rare or scrutinizes temporal features of the kinematics. Especially, the determination of spatial parameters is challenging due to common problems associated with integration of data from inertial sensors. This research is published by Bichler *et al.* (2012).

In the course of the development of an MC system for runners, this study aimed to determine stride parameters by foot-mounted IMUs and a smartphone GPS-receiver at a standard athlete running track. The stride parameters were length (SL), time (ST) and its inverse the rate (SR), stance/contact time (CT), flight time (FT), and speed (SS), which have been reported to be of importance for profiling the performance of athletes (Incalza, 2007). The behavior the stride parameters during running has yet not been fully substantiated, especially, how this knowledge might be employed to predict the performance and the risk of injury. From treadmill running and the evaluation of temporal features of time series, it became apparent that the analysis of variability and stability of the cyclic movement may allow for drawing conclusions upon the motor control system, and its relation to fatigue, performance and the risk of injury (Cusumano & Dingwell, 2013; Meardon *et al.*, 2011; Nakayama *et al.*, 2010; Dingwell & Cusumano, 2000; Bartlett *et al.*, 2007; Dingwell *et al.*, 2001). The behavior of stride parameters has been shown to differ between treadmill and over-ground (Schornstein, 2011; Dingwell *et al.*, 2001). Therefore, a measurement system providing running analyses with time series of temporal and spatial stride parameters may support answering to warranted research questions on this topic. Furthermore, such a system providing instantaneous stride parameters may be a basic tool for professional and amateur runners. One approach to on-line determination of stride parameters, especially SL, may be based on the pedestrian dead reckoning (PDR) algorithm (Fischer *et al.*, 2012).

The aims of this study were:

- Performing a qualitative evaluation of the envisioned stride-to-stride measurement system based on two foot-worn IMUs combined with a GPS-receiver on athletic tracks.

- Testing and evaluating of a vision-based reference measurement system, in order to design the actual evaluation study of the measurement system.

5.2 Method

5.2.1 The sensor setup

The Mobile Coaching system for runners (see previous chapter) was used primarily to gain the temporal GPS coordinates (sampling rate of $1\,Hz$). A smartphone (HTC Explorer, Android^{TM}, Athlete-Client) was mounted onto the subjects left upper arm with (a wide and) soft band. The Athlete-Client sent the data every $10\,s$ to a database server. A second smartphone (Apple iPhone 4) was fixated to the right arm. This unit functioned as metronome (zMetronome 2.0.2) and beeped at constant intervals. One IMU (Xsens^{TM}, XM-B-XB3, cable version) with a sampling frequency of $100\,Hz$ was firmly attached to each instep of the foot with a strap. The Xbus device, which is controlling the sensors, was fixated onto the waist with a belt. The data were streamed via a Bluetooth connection from the Xbus to a laptop, which was transported behind the subject by an operator on a bicycle.

5.2.2 Determination of stride parameters based on fused IMU/GPS

The *PDR* method uses at least one foot-mounted IMU, in order to track a walking path of a pedestrian. Various groups have reported good navigation results while walking, see e.g. Foxlin (2005), whereas at running speed this method is quite unstable. Nevertheless, for the envisioned scenario this method seems to be a viable approach, since the following additional information can be used: (1) GPS positioning and (2) the knowledge of the exact running track. Note, that the second demand constrains the application of the proposed system to tracks with exactly known coordinates. The proposed method is based on the algorithm described in Fischer *et al.* (2012) with some modifications due to the following considerations:

- The basic idea of most *PDR* algorithms is that the speed of the foot gets zero during the stance phase (Peruzzi *et al.*, 2011). This assumption can be used to correct the estimations for speed-, orientation-, and position-errors of the inertial navigation system (INS) by means of a Kalman Filter—this method is referred to as zero-updates (ZUPT). *PDR* algorithms depend on a stable detection of the stance phases though they are not very sensitive to low-frequent stance-phase insertion- or deletion-errors. The stance phase algorithm described in Fischer *et al.* (2012) was found to be very unstable for the running datasets on hand i.e., very frequent stance phase insertion- and deletion-errors. Since stance phases are also used to determine the start and stop of a single stride this method had to be improved.

- The entire *PDR* method needed to be extended, in order to deal with additional information sources e.g., GPS and track based heading information.

5.2.3 The Stance phase detection algorithm

The stance phase detection proposed by this work is optimized for runs and sprints. It is based on the following considerations: (1) there are only subtle changes of the step rate i.e., consecutive strides have similar lengths and duration. (2) Right and left strides are alternating. (3) There are no fast speed changes. (4) No backwards running. (5) During the stance phase the rotation rate of the foot is very low ($< 1\,rad^{-s}$). (6) Right before and after the stance phase the rotation rate of the foot is very high. (7) The heel strike results in a significant peak in the acceleration signal. The algorithm is thus designed as follows. The magnitude of the six-dimensional acceleration signal defined according to

$$a = \|[a_{x,right}, a_{y,right}, a_{z,right}, a_{x,left}, a_{y,left}, a_{z,left}] \tag{5.1}$$

$$\| \quad \| \ldots \text{Euclidian distance}$$

is filtered according to

$$a_f = w_1 \otimes a - w_2 \otimes a \qquad \otimes \ldots \text{convolution} \tag{5.2}$$

where w_1 is a 157-dimensional window representing a Butterworth low-pass filter with the cut-off frequency at $0.5\,Hz$ and w_2 is a 30-dimensional window representing a mean filter. The resulting signal a_f is approximately a sinus signal at step-frequency and a phase shift of approximately $-\pi/2$ with respect to signal a. Thus in between a minimum and a maximum of a_f exactly one initial contact can be found by a local maximum search in signal a within this time frame. The decision whether it is a right or a left initial contact is done by comparing the right and left gyroscope magnitude values at the first zero crossing of a_f after the heel strike (which lies within the stance phase). In case the right gyroscope value is low and the left is high at this point in time it is a right stance phase and vice versa. The stance phase can then be detected by searching for the first and the last gyroscope samples (of the respective foot) between two consecutive initial contacts that are below a certain threshold ϑ_g ($\vartheta_g = 1\,rad^{-s}$). For the dataset on hand, this strategy resulted in 100 % correct detection of stance events and thus 100 % correct stride events, which was evaluated by the approximately known stride time, which was constant for each single run, see fig. 5.1.

5.2.4 GPS processing

In order to combine both measurement systems, a common navigation frame had to be defined. Thus GPS results are converted from the geodetic reference frame into an ENU reference frame with the origin being close to the running track. This reference

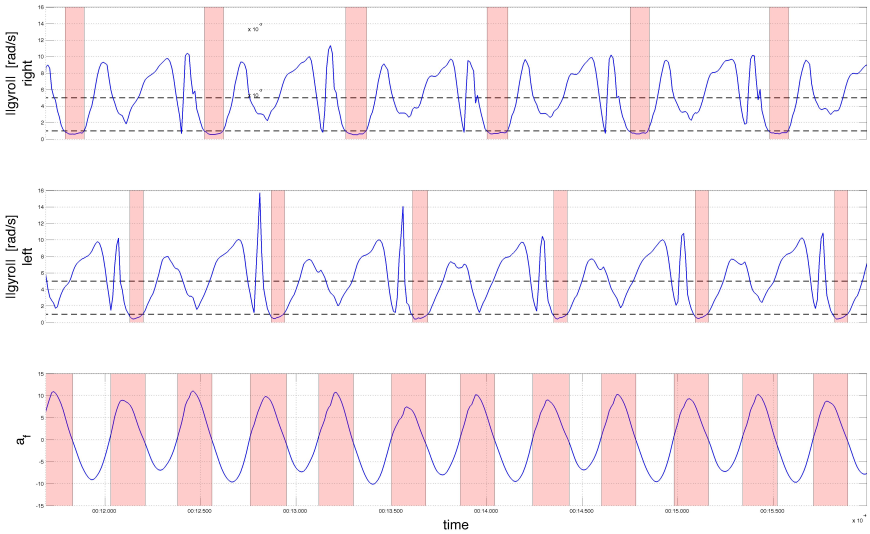


Figure 5.1: Zero update velocity assumption (ZUPT) during stance

frame serves as a common navigation frame. In addition, the exact running track was calculated from four GPS coordinates of the track: the corners of its rectangle exactly define a standardized running track, in this case these coordinates are ([latitude, longitude] in degree): $[[48.20428, 16.318647]$, $[48.2035272348501, 16.3185017269526]$, $[48.2034432366461, 16.3194757062384]$, $[48.2041960005762, 16.3196209933197]]$, see fig. 5.2. Furthermore, the type of the running track and the lane must be given, in this case type B (DIN18035-6) and most inner lane. The track is also converted into the navigation frame.

5.2.5 Sensor fusion

Various sensor fusion strategies are possible. The applied strategy assumes that the runner stays on one specific lane and thus the GPS coordinates are projected into this lane. From the result of the projection, the current expected yaw is calculated. The measurement model of the Kalman Filter proposed in Lee *et al.* (2010a) is extended to not only deal with the three velocity errors $[e_{vel,x}, e_{vel,z}, e_{vel,z}]^T$ but also with the yaw error. The error variance for yaw was set to $\sigma^2_{yaw} = 0.00052$.

Figure 5.2: GPS processing of the running scenario. The lane of the athletic track was adjusted with the knowledge of its geometrical model and the 4 GPS coordinates (X).

5.2.6 The reference system

Video data were collected using three cameras of the Canon MV 890 operating at 25 frames per second. The shutter speed was set to 1/1000. The cameras were placed in a line parallel $11.3\,m$ to the middle of the straight section of the $400\,m$ track. The distance between the cameras was $8.8\,m$. The height of the camera tripod was $0.2\,m$. To keep lens distortion low, an edge of about $1\,m$ of each camera was excluded. The overlapping zones of the cameras were marked to extract consecutive steps. The resolution of each camera was $960x720$ pixels. The focus was aligned up with the plane of the most inner lane. The kinematic data of the three (combined) cameras and consecutive laps were concatenated. Three operators digitized the video data using SIMI Motion®to obtain kinematic data of the strides by identifying the initial contact and the toe-off event. The intra class coefficient (ICC, two way random, single measure), the standard error of measurement (SEM), the coefficient of variation (CV), the mean and the standard deviation (STD) of the video data were calculated. In case of acceptable agreement, the average values of the three operators were taken. Finally, SL, ST, SR, CT, FT, and SS were calculated per stride.

5.2.7 Procedure

Three runs of two laps each at a different speed ($400\,m$) were recorded for each subject. At each stage, the speed was increased ($2.0\,m\,s^{-1}$; $2.5\,m\,s^{-1}$; $3.3\,m\,s^{-1}$). During each stage, however, the given speed was to be kept at a constant. Between each stage, there was a break of ten minutes. The $400\,m$ athlete track was marked every $10\,m$. According to the current speed, the subjects were asked to reach such a marker when the metronome beeped. On the day prior to the measurements, the subjects were allowed a trial run of three laps each to get used to running under these metronome conditions. Nine GPS satellites were available. The GPS sensor was switched on 10 minutes before the recording. The experiments were performed at a competitive all weather-running track, which had a polyurethane surface; 6 lanes, the most inner lane has got $400\,m$ in length. Three sport students (age=28.7 ± 1.5 years., height=$1.78 \pm 0.14\,m$, mass=$72.3 \pm 22.5\,kg$) took part in this investigation. All of them could ensure their health, were recreational athletes and were experienced in running with a routine of at least $10 \pm 5\,km$ per week.

5.2.8 Analyses

The determined stride parameters within the video surveyed area of the *IMU/GPS* approach were compared to the results of the video analysis. Data synchronization was done by time of day in milliseconds. The comparison of the two measurement systems was carried out by descriptive statistics, correlations (Pearson, *ICC*) and means of Bland-Altman plots.

5.3 Results

All operators evaluating the video data show a high agreement. This is expressed by an average *ICC* of 0.8, a *SEM* of $0.02\,m$ and measurement variability (CV) of $1.9\,\%$ in the entire data between the operators. Including GPS into the *PDR* implementation returned the following results, see tab. 5.1. An increase in speed raised the failure in all parameters. The mean of SL generated by *IMU/GPS* ($0.04\,m$) is smaller than the one gathered by video analysis (0.05). The mean of ST of *IMU/GPS* is approximately $0.1\,s$ smaller. ST and thus SR, CT, and FT of *IMU/GPS* had lower STDs and illustrated a more regular pattern than the parameters of the video analysis, but had some outliers.

The left and right SL results of *IMU/GPS* of all runs were highly correlated. That could mean that the algorithm works stably. The Pearson correlations of the two measurement systems were divers, mostly below 0.8 or even inversely proportional, e.g. for SL were from -0.37 to 0.54. The average *ICC* was 0.4. Bland-Altman plots show

Table 5.1: Differences of stride parameters between *IMU/GPS* and reference system

Stride parameter	Speed 1 $= 2\,m\,s^{-1}$	Speed 2 $= 2.5\,m\,s^{-1}$	Speed 3 $= 3\,m\,s^{-1}$
SL	$0.00 \pm 0.01\,m$	$0.07 \pm 0.11\,m$	$0.14 \pm 0.54\,m$
SR	$0.00 \pm 0.07\,Hz$	$0.00 \pm 0.13\,Hz$	$0.00 \pm 0.08\,Hz$
ST	$0.00 \pm 0.04\,s$	$0.00 \pm 0.05\,s$	$0.00 \pm 0.03\,s$
CT	$0.14 \pm 0.04\,s$	$0.12 \pm 0.04\,s$	$0.08 \pm 0.03\,s$
FT	$-0.14 \pm 0.04\,s$	$-0.12 \pm 0.06\,s$	$-0.09 \pm 0.03\,s$
SS	$0.06 \pm 0.13\,m\,s^{-1}$	$0.016 \pm 0.17\,m\,s^{-1}$	$-2.6318.15\ s^{-1}$

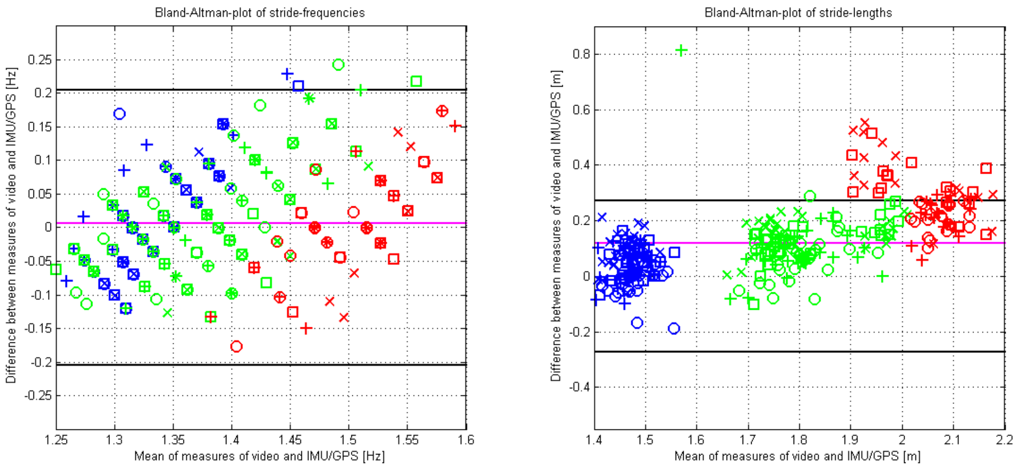
that differences were mostly within the 95 % limits of agreement; see fig 5.3. Both measurements have low variability and differences. High STD could just be detected in the most sensitive parameter stride-length at higher speeds.

5.4 Discussion

The results of the correlations dismiss the expectations of the reliability descriptive statistics initially suggested. From this study, the perfect match could not be established. In this case, paired correlation may be useless, even if differences are small. It might be that this video analysis was too weak to be used as the reference system, due to the higher standard deviation in comparison to *IMU/GPS*. However, it was noted outliers were present in *IMU/GPS* results, which deteriorate the overall results. The high mean values of contact-time and flight-time are due to the different definitions of the stance phase for both measurement systems i.e., CT (video) versus no-motion-time (*IMU/GPS*). Further improvement of the precision of the parameter SL is even more important than a systematic error would decrease accuracy by under- or overestimating the results. This is important for the analysis of variability.

5.5 Conclusion

This measurement method based on *IMU/GPS* promises to have a great potential for instantaneous, high precision determination of stride parameters in running. The qualitative analyses came up with realistic values for all stride parameters. So far, it worked well for slow to medium speeds. The speeds for elite runners should be drawn into consideration in the future. Reliability check is an outstanding demand, which



(a) (b)

Figure 5.3: Bland-Altman Plots a) Stride rate b) Stride length. Blue. . . speed 1, green. . . speed 2, red. . . speed 3, +. . . left foot lap 1, x. . . right foot lap 1, o. . . right foot. . . lap1, square. . . right foot. . . lap 2; Magenta line. . . mean of differences, black lines. . . $2 \cdot STD$ of differences

has to be reached with a highly improved reference system. Movement variability or even stability in running could be investigated in the meaning of a functional concept. This encourages performing further studies to high-precision *IMU/GPS*-based stride-parameter determination.

6

Behavior of stride kinematics in a 5 km time-trial

6.1 Introduction

This study examined the stride kinematics and its variability in middle-distance running with the intention to provide the project of the Mobile Coaching (MC) system for runners with knowledge to exploit the stride parameters for predicting performance, reducing the risk of injury and estimating the effort of runners. Due to restrictions of measurements in over-ground running, there are only few investigations available. For example, Belli *et al.* (1995) suggested to consider at least 30 strides to analyze the behavior such as variability in the stride pattern. Moreover, studies in over-ground running disagree with the results of the treadmill based studies. It appears that the stride length (SL) on treadmills increases as runners fatigue (Hunter & Smith, 2007; Candau *et al.*, 1998; Mizrahi *et al.*, 2000), whereas in over-ground running SL tends to decrease (Hanley & Smith, 2009; Chan-Roper *et al.*, 2012; Kyroelaeinen *et al.*, 2000; Buckalew *et al.*, 1985). In contrast, the stride rate (SR) was not consistent in the long distances, though it increased in the studies of Kyroelaeinen *et al.* (2000) and Buckalew *et al.* (1985) conflicting with Chan-Roper *et al.* (2012). Obviously, adjustments occur under fatiguing conditions.

In middle-distance running, Hanley & Smith (2009) is the only study that depicted the time course of SL and SR in a $5\,km$ time-trial. Nevertheless, about two to three strides were captured three times over the whole race by a video analysis, and therefore, allowed for general statements of the overall time course. However, the interplay between SL and SR remained covered. SR is assumed to be an indicator of the neuromuscular potential. While the efficiency of the musculotendineous apparatus were diminishing, several adjustments had been recognized. For example, the contact time (CT) and vertical acceleration had increased. The overall decrease in SR were only small compared to the greater decrease of SL. Nevertheless, SR appears to play a crucial role for the efficiency and even small changes might suggest a lowered motor potential. To date, different schools of running styles controversially debate the optimal SL (Fletcher *et al.*, 2010; Elliott & Blanksby, 1979). However, in consensus, over-striding is associated with a higher metabolic cost and a greater risk of injury (Diebal *et al.*, 2011; Romanov & Fletcher, 2007; Cavanagh & Williams, 1982). As SR remains quite stable, SL correlates with the performance. Therefore, SL should reach the highest value, when efficiency is still guaranteed and there is a low risk of injury. From elite runners, the optimum was derived by e.g., about from 100 % to 115 % of the body height (Cavanagh *et al.*, 1985; Scholich, 1978). During the race, the conditions and constraints of an optimal movement pattern might change. The movement task is desired to be solved within the spectrum of possibilities according to the minimum energy hypotheses (Miller *et al.*, 2012) predominantly in the first half of the race, in order to gain high performance. The shift towards compensation, which should establishes a stable movement, is reached through a certain amount of variability in the kinetic chain. Although it is not clear whether or not variability is initiated or negotiated by the neuromotor control structures, it can allow for changes in the environment and prevent overload of the involved tissue within the kinetic chain according to the variability-overuse injury hypothesis (Bartlett *et al.*, 2007; Wheat, 1985)—suggesting an increase of variability under elevated fatigue. Fatigue was considered as the "inability to sustain a target work rate" (Gates & Dingwell, 2011) and having influence on the motor potential. In running, the overall (average) variability of the running speed ensuring efficiency is considered to be of approx. 3 % when expressed by the coefficient of variation (Billat, 2001). Cottin *et al.* (2002) proved that the variations in the running speed neither raise with fatigue nor cause fatigue. The overall movement remains stable, and hence, possible adjustments may occur in SL and SR. The variability of these single stride parameters might be indicators of changes in the conditions of movement and further relate to the state of the runner.

Derived from some elite middle-distance runners, Saziorski *et al.* (1987) proposed a model, which predicts the behavior of the running speed, SL and SR when fatigue sets on. Herein, the adjustments arise in three phases, see chap. 2.1. To the knowledge of the author, this model has yet not been evaluated for recreational runners. Moreover, there were no information available that could describe the variability of the stride parameters under fatiguing conditions. Therefore, the goal was to identify kinematic markers of fatigue in a 5 km full exhaustive run by illustrating the time course and its variability of the stride parameter. The evaluation of fatigue and the related motor potential may support the training while determining the load on the athlete, and thus helps to improve performance, select appropriate intensities and/or avoid fatigue-related injuries. SL was hypothesized ($H1$) to correlate with performance stronger than SR and decreases with fatigue ($H2$). SR is assumed to provide a compensatory strategy for keeping the speed in the second third of the race ($H3$). Both SL and SR are expected to decrease in the last third of the race ($H4$). Experienced but non-competitive runners are hypothesized to have a coefficient of variation greater than $3\,\%$ in all stride parameters. As fatigue proliferates, the variability expressed by the standard deviation would increase ($H6$) causing also a change in the stride pattern e.g., SL decreases and SR compensates. Nevertheless, both SL and SR as a function of stride speed (SS) were expected to display a monotonic behavior ($H7$) due to visual observations that the running style does not change within a $5\,km$ time-trial.

6.2 Method

6.2.1 Subjects

Eleven male recreational runners gave written informed consent for this study. The experimental protocol was approved by the Ethics Commission of the University of Vienna. In the first experimental session, they were asked to complete a questionnaire about their age, health status, training habits, race records, shoe inserts, foot type, chronic diseases, medication and history of (orthopedic) injuries with focus on the lower extremities. The runners were healthy and had no biomechanical abnormalities. Further, they were familiarized with the experimental protocol and with the rated perceived exertion scale (RPE) with 15 levels according to Borg (1982) (Borg-15). This session also consisted of some anthropometric measures. The body weight and fat content were scaled with a glass body fat monitor (UM 072, Tanita®, Health Equipment H.K. LTD, Kowloon, Hong Kong). The mean subject age, height, weight, and leg length (LL) were 30 ± 11 years, $1.82 \pm 0.06\,m$, $76.8 \pm 11\,kg$, $0.95 \pm 0.03\,m$, respectively. The body composition was characterized by the body mass index (BMI) and the fat content amounting to 23.11 ± 2.4 and $15 \pm 3\,\%$, respectively. On average, they had a weekly training volume of $7\,km$. A cardiologist examined their medical qualification for high intensive cardiovascular exercises. The runners stated to be non-fatigued and rested as at least one week prior to the time-trial they volunteered in an continuous incremental stage test on a motor driven treadmill (ERGO ELG2, Woodway®, Waukesha, USA) to full exhaustion. This test included a continuous heart rate measurement operating at $1\,Hz$ (Polar®S725, Polar Electro, Finland) and respiratory gas analysis (Oxycon

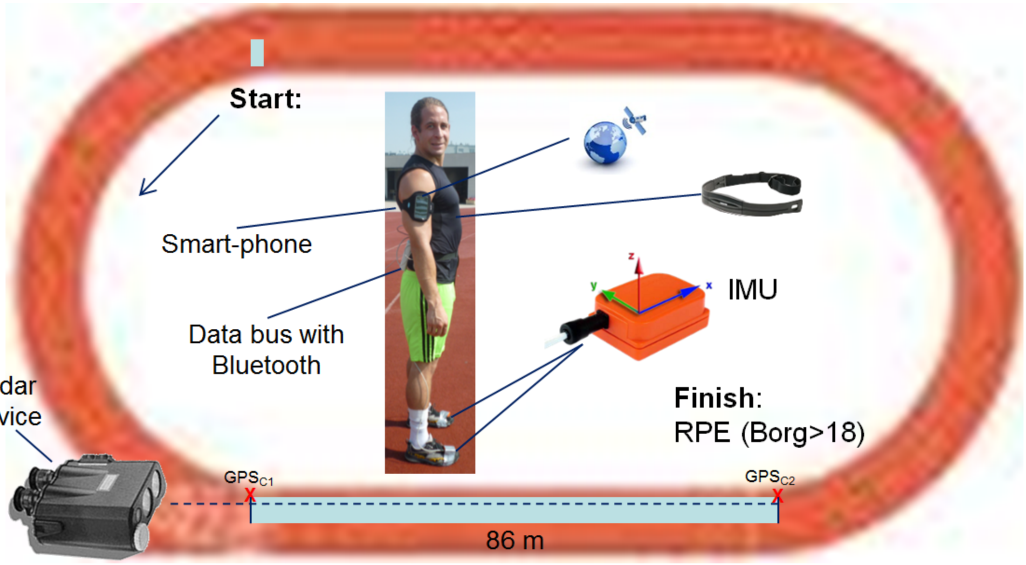


Figure 6.1: Experimental set-up: The race was conducted on an athletic track. The runners were equipped with a smart-phone collecting GPS and heart rate data and with an inertial sensor at each foot. On a long stretch their horizontal displacement over time were measured by a radar device.

Pro^{TM}, CareFusion GmbH, Germany), in order to assess their running performance and economy. The inclination of the treadmill was $1\,\%$, which should compensate for the headwind of over-ground running (Jones & Doust, 1996). The running speed of the first stage was $6\,km^{-h}$. Each stage lasted one minute. Dependent on their race records, the inclination in speed per stage was $0.8\,km\,h^{-1}$ or $1\,km\,h^{-1}$. The mean $V02/kg$ at the anaerobic threshold ($T2$) and the maximal oxygen uptake per kg of the body weight ($V0_2/kg$) were $46 \pm 4.5\,ml/min/kg$ and $53 \pm 6.4\,ml/min/kg$, respectively. At the anaerobic threshold, the mean heart rate (HR_{T2}) and the running speed (V_{T2}) were $180 \pm 10\,bpm$ and $13.8 \pm 1,5\,km\,h^{-1}$, respectively. The mean maximal heart rate (HR_{max}) and maximal running speed (V_{max}) were $193 \pm 8\,bpm$ and $15.8 \pm 1.9\,km\,h^{-1}$. Shoes were not prescribed, neither for the performance diagnostic nor for the time-trial.

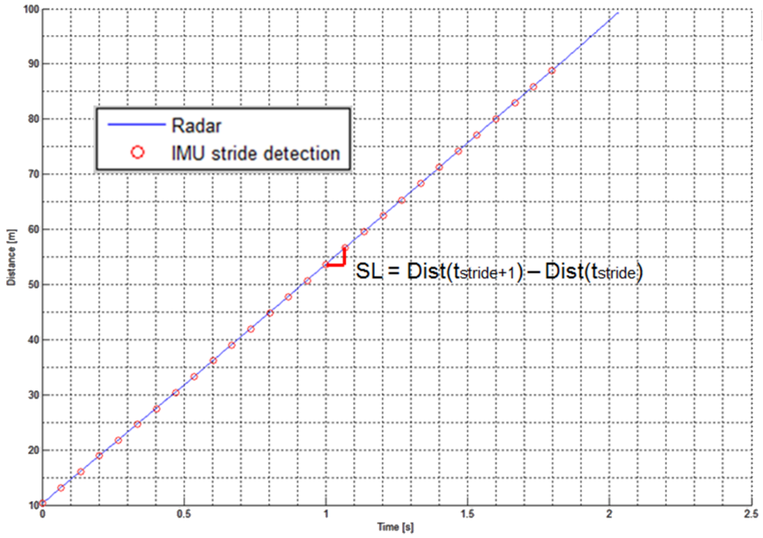
6.2.2 Experimental protocol

In order to limit fatigue, subjects were asked to rest and abandon training at least two days before the race. The time-trial took place on the inner lane of a standard athletic oval track (type B, DIN18035-6) being 400 m in length, see fig. 6.1. After the runner were equipped with the measurement devices, they were informed once again about the procedure, especially the RPE. A 15 min lasting warming-up should prepare them for the high intensive run by including three short intensive bouts above 80% of their HR_{max}. At the end of the warming-up, two laps at a self-selected comfortable slow speed were recorded, in order to ensure the cooperativeness of the measurement devices. Races were only conducted if there was no or very light wind. Temperature was $26 \pm 2.6°$. Based on the results of the performance diagnostic, they were motivated to achieve a race duration at least when running at 3 % above their V_{T2}. No feedback was provided to the runner during the race. The start line was near by a corner of the inner field. From there on, twelve and a half laps were counted. The 100 m finish line at the opposite diagonal corner of the inner field was used as the finish line of the race. A whistle gave the signal for the start. At the finish, they had to rate their perceived exertion level immediately. They were shown a sheet of paper depicting the Borg-15 scale (German version according to Lollgen (2004) and then replied orally or pointed to the referred level on the paper. Full-exhaustion was confirmed by a Borg-15 level of at least 18 and a running speed by abiding by at least 3 % above V_{T2} .

6.2.3 Data acquisition

The MC system for runners proposed in chap. 4 collected the data of the heart rate and the GPS position at a smart-phone attached on the upper right arm with a strap. The heart rate sensor (HRM1, Garmin Ltd., USA) was placed on the rib cage with a chest-strap. The GPS receiver was a standard, low-grade sensor (GPS class 0). Both GPS and heart rate sensor operated on a sample rate of 1 Hz. In an interval of 10 s, the client software on the smart-phone sent the data to a databases server. An inertial measurement unit (IMU) (XSens^{TM}, XM-B-XB3, cable version) operating at 100 Hz was attached to each hind foot and fixated with a tape. The IMUs were connected to a control unit (Xbus) via cables. This unit was fixated onto the waist with a belt. From there, the sensor data were transmitted via a Bluetooth connection to a laptop, which was transported behind the subject by an operator on a bicycle. The cables connecting the sensors and the control unit were carried under the shorts of the runner and strapped at the calf with bandage material (Peha-haft^{TM}), in order to impede the oscillation of the cables or even the extraction from the sensors. A radar measurement device (Laveg, LDM 300 C, Jenoptik, Jena, Germany) captured the horizontal displacement over time of the runners on the straight being $86\,m$ in length ending at the finish line for each lap i.e., the radar distance equaled $22\,\%$ of the strides of each lap and the whole race, respectively. The width of a lane was standard i.e., $1.22\,m$. Runners used to run close to the inner boundary. Therefore, the projection of the laser was adjusted to $0.45\,m$ in parallel from the inner boundary of the lane. The manufacturer explicitly suggested this radar device for the application in sports. Its sample rate was $100\,Hz$. The precision of measurement was $0.1\,m\,s^{-1}$ until $10.0\,m\,s^{-1}$ if the coefficient of the laser reflection was greater than $20\,\%$. Therefore, the runners had to wear a white T-shirt. The extended *PDR* method (see chap. 5) was used to determine SR (beginning of a stance phase e.g., Mariani *et al.* (2010)). As a function of time, both SR and the distance of the radar measurement were fused, in order to determine time series of SL. Therefore, the synchronized time lines of both parameters were paired. SL equaled the difference of two consecutive strides, see fig. 6.2. This measurement method was named *IMU/RADAR*. The radar measurement device was chosen over the *PDR*-method due to its higher precision over the range of running speeds in middle-distance running. The measure of the distance was not smoothed. Each radar measurement was visually observed, in order to detect possible failures e.g., when the runner was out of the radar scope for a short time. Ten of 143 ($= 13 \cdot 11$) measurements with failures were excluded.

The start and the finish line of the radar measurement were referenced by GPS coordinates as these lines crossed the inner boundary of the lane (determined with http://www.wien.gv.at/stadtplan/), see fig. 6.1. The synchronization of each lap was set to the time point as the runner put one foot over the start line of the radar measurements. Therefore, a vector was created from the GPS coordinates before and after the start line by projecting both coordinates into the inner boundary of the (straight) lane. The length of the distance from the start along the vector to the projection of the start line was used to determine this time point with linear interpolation. At the radar device, an operator noted the foot, which crossed the start line of the radar measurements at first.



(a)

Figure 6.2: *SL* determination by fusing data collected from the radar and the IMUs

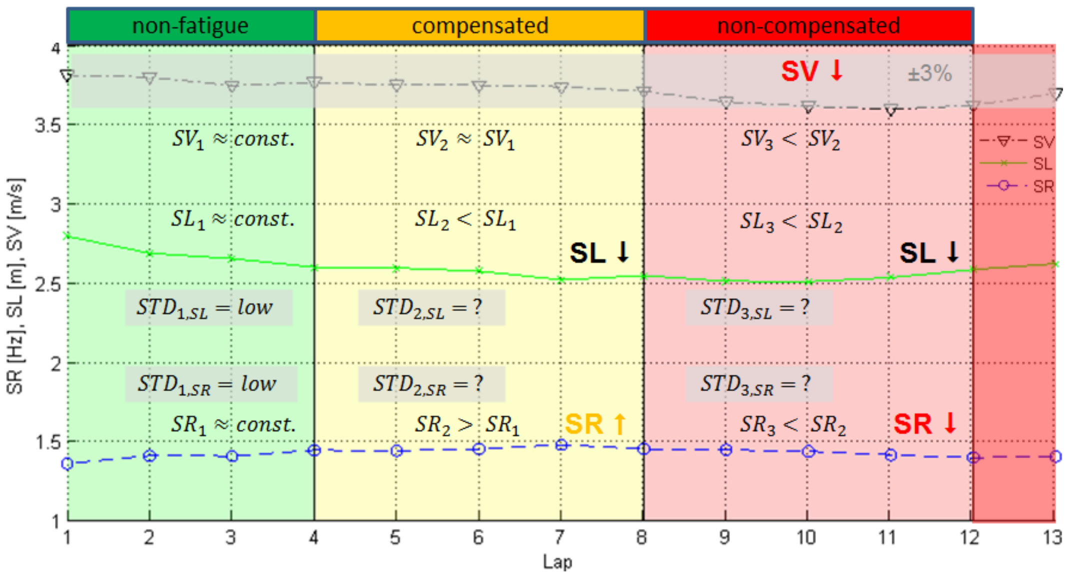


Figure 6.3: Kinematic model according to Saziorski *et al.* (1987)

6.2.4 Data analysis and reference model

The model of the stride parameters according to Saziorski *et al.* (1987) is characterized by three phases, which are described as 1) non-fatigue, 2) compensated and 3) non-compensated, see fig. 6.3. The first phase shows no signs of fatigue but a stable relation between SL and SR expressed by their low variability. In phase two, SL decreases, but SR compensates, and therefore SS is nearly the same as in phase one. This neuromuscular adaptation is assumed to occur unconsciously. Finally, both parameters decrease, and so does SS. The model did not provide information about the variability of the stride parameters in phase two and three. Variability for all stride parameters was expressed as the standard deviation (STD) and the coefficient of variation (CV). In this case, phase one included laps one to four, phase two five to eight and phase three nine to twelve. The final lap was considered separately. An analysis of variance ($ANOVA$) with a significance level of 0.05 and the factor lap was calculated for the stride parameters, in order to examine the presence of adjustments. The post hoc test Bonferroni was used for multiple pairwise and intra-individual comparisons of the laps. Descriptive statistics were calculated for each race and lap. Further analysis operated on the average data for all stride parameters, in order to generalize the trends in the parameters. Pearson's product moment correlation coefficient was used to indicate the strength of the relation between SS and SL and further between SS and SR. Significance level was set to 0.05. The monotonic trend in SL and SR as a function of SS were tested as the assorting the time series by SS ascending. Linear, quadratic, and spline regressions were applied for each phase, stride parameter and its variability. Based on these results kinematic markers indicating stages of fatigue with relation to performance and the injury risk were established on the average data. The time course of all parameter were also compared to the 17 competitive runners (age 26 ± 4 years, height $1.72 \pm 0.6\,m$, mass 60 ± 7) of the study of Hanley & Smith (2009).

6.3 Results

Mean RPE, HR and running speed were 18.7 ± 1.4, $180 \pm 8\,bpm$ and $12.4 \pm 1.8 km\,h^{-1}$, respectively. The mean overall SS, SL and SR were $3.58 \pm 0.30\,m\,s^{-1}$, $2.63 \pm 0.15\,m$ and $1.36 \pm 0.03\,Hz$, respectively. The mean STD of the overall SS, SL and SR were $0.55 \pm 0.08\,m\,s^{-1}$, $0.07 \pm 0.03\,m$ and $0.024 \pm 0.011\,Hz$. The mean CV of SS, SL and

Table 6.1: Average stride parameters

Phase	$SS\,[m\,s^{-1}]$	$SL\,[m]$	$SR\,[Hz]$
1	3.67 ± 0.10 (2.7 %)	2.68 ± 0.08 (2.9 %)	1.369 ± 0.003 (0.2 %)
2	3.45 ± 0.05 (1.4 %)	2.56 ± 0.03 (1.2 %)	1.349 ± 0.009 (0.7 %)
3	3.40 ± 0.05 (1.5 %)	2.53 ± 0.04 (1.6 %)	1.343 ± 0.006 (0.4 %)
4	3.98 ± 0.69 (17.3 %)	2.82 ± 0.34 (12.1 %)	1.401 ± 0.074 (5.3 %)
overall	3.58 ± 0.30 (8.4 %)	2.63 ± 0.15 (5.7 %)	1.36 ± 0.03 (2.2 %)

Table 6.2: Average stride parameters according to Hanley & Smith (2009)

Phase	$SS\,[m^{-s}]$	$SL\,[m]$	$SR\,[Hz]$
1	5.92 ± 0.27 (0.5 %)	3.68 ± 0.14 (0.4 %)	1.61 ± 0.07 (0.4 %)
2	5.70 ± 0.36 (0.6 %)	3.60 ± 0.20 (0.6 %)	1.58 ± 0.06 (0.4 %)
3	5.62 ± 0.41 (0.7 %)	3.58 ± 0.22 (0.6 %)	1.57 ± 0.06 (0.4 %)
overall	5.78 ± 0.33 (0.6 %)	$3.64 \pm 0,18$ (0.5 %)	1.59 ± 0.06 (0.4 %)

SR was $9.21 \pm 3.67\,\%$, $6.93 \pm 2.87\,\%$ and $2.67 \pm 1.41\,\%$. Therefore, $H5$ was accepted at least for SS and SL. Subjects exhibited significant changes in all stride parameters over the time-trial. The first and the last lap were similar but differed mostly from all others laps. The post hoc test revealed that there were similar laps. In the majority of the subjects, there were two to three laps with a similar mean in sequence. Mean and standard deviations of the average data is given in tab. 6.1 and the mean and standard deviation of STD across the laps of runners in tab. 6.3. All stride parameter are smaller in the recreational than in the competitive runners. SL decreased until phase three, then increased and was highest in the end. Contrary, the competitive runners decreased SL gradually. CV is greater in the recreational than in the competitive runners. SR was on average the same as the competitive runners demonstrated during the first three phases. The mean STD displayed a high CV in all stride parameters and phases. In the final phase, CV of all stride parameters is higher compared to all phases in before. CV is remarkably greater in SS and SR than in SL.

The linear regressions are presented in the plots, see fig. 6.5. SS and SL were fitted with a quadratic regressions: $SS = 0.0046x^2 - 0.0626x + 1.0763$ and $SL = 0.0035x^2 - 0.0485x + 1.0545$, respectively, where x denotes the lap. The significant correlations of SS to SL and SR were 0.99 and 0.96, respectively, thus $H1$ was accepted. Nevertheless, SR was also very strongly correlated to performance. Moreover, both SL and SR

Table 6.3: Mean STD of stride parameters

Phase	$SS\,[m\,s^{-1}]$	$SL\,[m]$	$SR\,[Hz]$
1	0.08 ± 0.02 (25.0 %)	0.07 ± 0.01 (14.3 %)	0.022 ± 0.002 (9.0 %)
2	0.07 ± 0.02 (25.6 %)	0.07 ± 0.01 (14.3 %)	0.024 ± 0.003 (12.5 %)
3	0.07 ± 0.01 (14.3 %)	0.06 ± 0.01 (16.7 %)	0.021 ± 0.001 (4.8 %)
4	0.15 ± 0.10 (66.7 %)	0.09 ± 0.03 (33.3 %)	0.032 ± 0.015 (46.9 %)
overall	0.09 ± 0.04 (32.9 %)	0.07 ± 0.01 (19.7 %)	0.02 ± 0.01 (73.0 %)

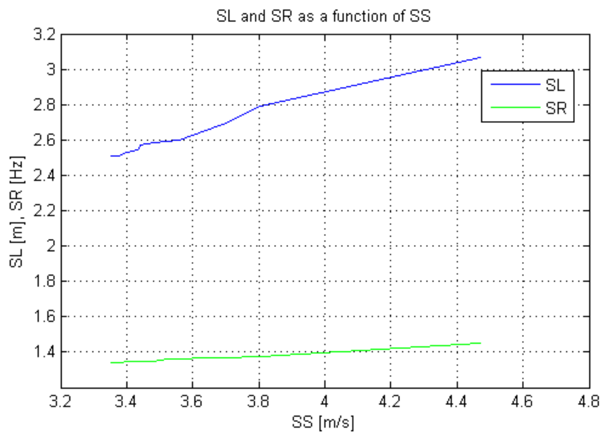


Figure 6.4: SL and SR as a function of SS

exposed a strictly monotonic function of SS, see fig. 6.4; hence $H7$ had to be accepted. Metabolic and neuromuscular fatigue was assumed to increase during the race. The performance and mainly SL decreased until the end of phase two, then increased. This decrease might be referred to an elevated fatigue, however, for a while these runners under high fatigue were able to increase their SL. In general, $H2$ was rejected. SR did not compensate in the second third but in the first, therefore $H3$ was rejected. Regarding the descriptive statistics, both SL and SR decreased, therefore $H4$ might be accepted. However, the linear regression of the SL revealed a slight increase towards the end of phase three. The descriptive statistics also negated an increase in the variability as $H6$ claimes. In recreational runners, the behavior of the stride parameters were not fully in accordance with the predictions of the model according to (Saziorski *et al.*, 1987).

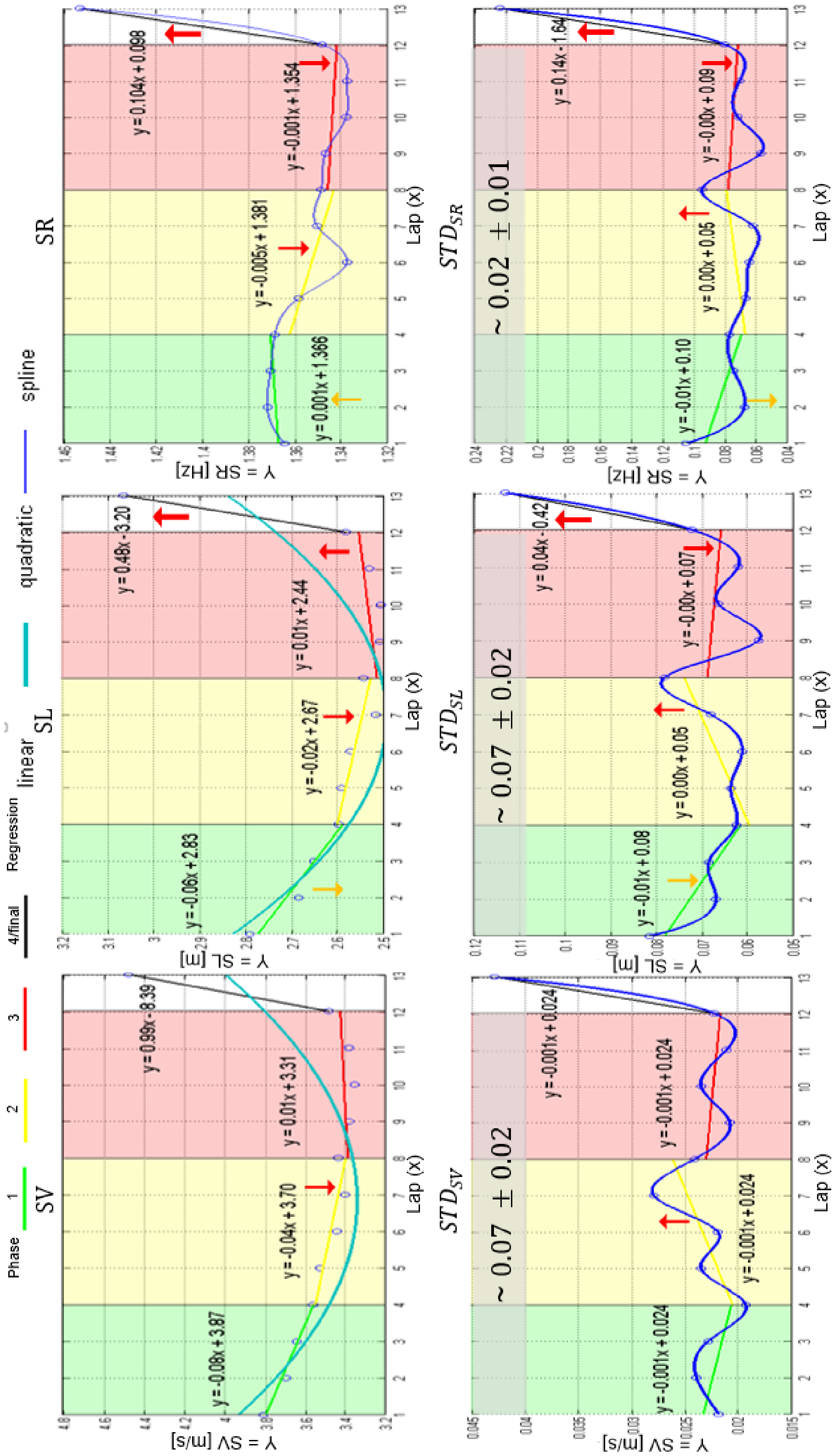


Figure 6.5: Time course and variability of stride parameters

6.4 Discussions

This study was concerned with the behavior of the stride parameters and its variability over the time course of a 5 km time-trial. The measurement method comprising a radar and IMU allowed for obtaining at least 24 strides (48 steps) for each lap in over-ground running and therefore measures of variability (mean STD) could be calculated. A model derived from elite runners was used to assess recreational runners or to validate its usability for the establishment of kinematic markers, respectively. This study confirmed adjustments in the stride pattern and extended their aspects regarding variability. The variability in the stride parameters was higher compared to athletic runners—as expected. Both competitive and recreational runners were relatively homogeneous in their stature. Nevertheless, the competitive runners were similar in their stride patterns over the race but were contrasted by the recreational runners. There was no comparison available for a series of strides, but CV of the recreational runners in this study suggested that the stride pattern demonstrated a high variability, and therefore might be characterized as unstable. SL is highly correlated with performance. It appeared that the adjustment of SL from the beginning influenced the stability and thus the efficiency of the stride patterns. Although SR displayed on average the same dimension as in the competitive runners, little fluctuations impacted the overall stability. Especially at the end, the variability raised—perhaps representing very unstable movements when compared to the competitive runners. This might endorse the variability-overuse injury hypothesis. SS and SL might be approximated by an U-shaped function, which has been shown to play an important role in gait analysis regarding stability. The minimum of this function is in both parameters in the middle of the run—arguing that high SL could be only achieved by high volition. The neuromuscular control might have dragged the runner to adjust towards a stable or comfortable stride pattern. This suggests that the speed at the beginning was adversely chosen.

The variability expressed by descriptive statistics did not change between the phases; therefore the linear and spline approximations were used for a qualitative interpretation. The spline approximation revealed a wave-like pattern, especially in the curves of the variability. Assuming that these waves would represent bursts of high neuromuscular stress, it would mean that there is a temporal compensation in the wave troughs i.e, when variability is smallest. It was expected from the model according to Saziorski *et al.* (1987) that there is a stable relation between SL and SR in phase one, whereby stable was referred to similar mean values between the laps. In phase one, the linear regressions illustrated a downward trend. SS decreased, also SL—gradually until the end of phase two, and in phase one SR compensated, not in phase two as expected. Probably, SR compensated very early due to the adversely chosen SS (and SL) at the beginning. The runners though might have shown the first signs of fatigue in phase one. The high variability of SL in the beginning could characterize recreational runners. The first wave of SR's variability decayed while it was passing into phase two. The runners might have stabilized or somehow recovered. In phase two, the runners tended to get more and more fatigued but unconsciously. That can be seen in a nearly monotone decrease in SS and in the increase in variability. The runners optimized, attempted to adapt to the elevated fatigue. SL and SR decreased further in phase two, that would have occurred in phase three according to the model. Regarding *RPE*, fatigue increased progressively. Variability increased near before phase three. Although fatigue is assumed to be higher in phase three than ever before, variability in SR oscillated in a smaller range and tended to decrease in SR and SL. Under these high fatigued conditions, SR decreased; but SL increased slightly. It can be speculated that the runners got aware of being near to the end and tried to enforce an increase in their performance. The variability of the SL decreased to this time point. This suggested that phase three is richer in the consciousness exposure than the previous phase. Further striding is associated with a higher energy consumption compared to higher stride rates. This may mean that a decreased neuromuscular mechanism would have caused this stride pattern. Strictly speaking, the stretching-shortening cycle was reduced, and therefore they had to stride further. This means that the energy expenditure increases, because this technique is very inefficient. Striding further under high stress condition can lead to an over-striding which is associated with a higher risk of injury. The high fatigue might be represented by perturbations i.e., the wave-like pattern of *STD*. Although there was a decrease in the variability near the end of phase three, at the end spurt, the stride pattern changed. An increase in performance could be achieved by increasing all stride parameters and its variability. The high increase in the variability might portray the high fatigue.

Phase	SP	Fatigue	Performance	Injury risk
-	$STD_{SL,SR} = const.$ (small waves)	low	stable, but low	low ?
1 (2)	$SL\downarrow$, $SR\uparrow$	moderate	$\downarrow$	-
2 (3)	$SL\downarrow$, $SR\downarrow$, $STD_{SL,SR}\uparrow$	high	$\downarrow\downarrow$	-
3 (4)	$SL\uparrow$, $SR\downarrow$, $SV\uparrow$	very high	$\uparrow$	($\uparrow$?)
final	$SL\uparrow$, $SR\uparrow$, $STD_{SL,SR}\uparrow$, $SV\uparrow$	severe	$\uparrow$	
2,...	$STD_{SL,SR}\uparrow$ (dynamic waves)	$\uparrow$?		$\uparrow$?
1-4	SV		±3%	-

Figure 6.6: Kinematic markers

6.5 Conclusions

Taken collectively, the behavior of the stride parameters was not fully in agreement with the prediction of the model according to Saziorski *et al.* (1987). Nevertheless, it may be applied as a goal in training for optimization towards its predictions in elite runners. With the help of this model and the analysis of variability, it seems to be possible to establish kinematic markers. Based on these observations, a new model of kinematic markers was developed to relate fatigue, performance and the injury risk, see fig. 6.6. The variability of SL in the recreational runners was highest compared to SR and was directly correlated with performance. Runners displaying these symptoms are suggested to rely on training methods improving their stride ability. The compensatory strategy could be clearly seen although it happened in an earlier phase. As a recommendation these runner should slow down if the goal would be to train in a moderate training zone or to better budget the resources over a time-trial. High performance, but a decrease in SR, and an increase in SL was concluded to mark a very high level of fatigue and a low motor potential. The analysis of variability may promote criteria in a biomechanical assessment e.g., when runners are placed at unstable phases representing a link to the risk of injury. The changes observed may represent different levels of neuromuscular stress. Therefore, SL and SR are promising parameters to determine training load or the state of the runner, respectively, and thereby support feedback provision to runners during training. Further research is should prove the relation between variability and the risk of injury in over-ground running.

7

Indicators of fatigue in a 5 km time-trial

7.1 Introduction

The last chapter proved the time course and the variability of the stride parameters in a middle-distance time-trial for their usability in a biomechanical assessment. It was confirmed that fatigue emerges in the stride pattern, and thus the load on the athlete can be estimated by kinematic markers. Moreover, this estimation allowed to draw conclusions upon performance and the injury risk. In the literature more common parameters have been found, which to date have not been used in a biomechanical assessment, but were of interest in investigations of human running. The purpose of this study was to examine further parameters as indicators, which may support a biomechanical assessment based on inertial sensors. Due to the diminishing of the neuromuscular efficiency, the contact time (CT) or the (force) loading rate (*1/CT*) is known to increase or decrease, respectively, during a full-exhausting run, see e.g, Kram & Taylor (1990). CT is the starting point for the temporal based analysis of strides. In this study, it was hypothesized that CT would increase ($H12$), likewise for the stride speed (SS) normalized CT, CTn ($H16$). During fatigue and its associated decrease in stride length (SL), it was thought that flight time (FT) and its for SS normalized parameter (FTn) would decrease ($H13$, $H17$). The behavior of the duty factor (*CT/ST*) was examined with the expectation of its increase ($H14$), while the ratio of the (FT) to ST (*FT/ST*) ($H15$) was expected to decrease. The heart rate-to-contact time index (*HR:1/CT*) introduced by Oliver & Stembridge (2011) combines physiological and biomechanical measures. If this index could provide further indications for the estimation of fatigue of the runner, then the MC system for runners may take advantage of it. According to its developers, this index was hypothesized to increase with fatigue ($H11$) due to a very high heart rate (HR) at least above the anaerobic threshold ($T2$) and an increase in the CT. From the last chapter, it appeared that the monotonic behavior in both SL and SR as a function of SS was present also under fatigue. In this context, it was hypothesized that the running relation ($RL = SL/SR$) in contrast to the walking relation (Terrier & Schutz, 2003) would decrease or at least would not remain constant with fatigue ($H10$)—based on the assumption that SL would decrease with a temporary compensation of SR. Although Billat *et al.* (2003) and Cottin *et al.* (2002) uttered that the distribution of the running speed within about 3 % does not remarkably influence fatigue and vice versa, the variation in all the stride pattern in the recreational runners were about 10%, and therefore, might have caused inappropriate energy expenditure. In order to examine the influence of the speed on the stride pattern, the for SS normalized stride parameter (SLn) and (SRn) were compared to their originals. It was hypothesized that SLn and SRn would behave as SL and SR ($H8$). Furthermore, it was made a draft upon on the stride rate index (SRI), which integrates SS with regard to the time. The results of the last chapter and the literature suggest that recreational runners regulate their changes in speed mostly through changes in SL. Therefore, the SRI index was further supposed to be low in the beginning (SL dominates); to increase in the middle due to the compensatory strategy of SR, and at the end would increase ($H9$). In recreational runners, changes in speed were suggested to be regulated by changes in SL. The last purpose of this study was to investigate whether or not recreational runners expose an optimal SL and an optimal SR. Did they over-stride and if—how did fatigue influence the over-striding? Herein, it was hypothesized that they did over-striding ($H18$). It was assumed that recreational runners would have a lower than optimal SR ($H19$).

7.2 Method

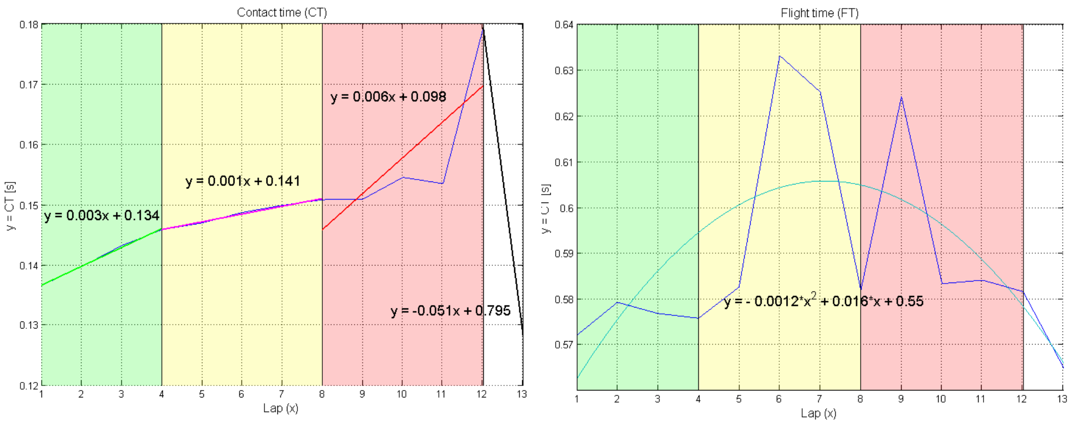
This study was part of the same experiment introduced in the last chapter. The temporal parameters were obtained with an inertial sensor unit (IMU) placed at each hind foot, whereas the spatial parameters were determined by fusing the temporal information with the radar measurements. Although the data of the temporal stride parameters were available for the whole run, this study focused on the data gathered on the thirteen laps of the $86\,m$ radar distance, in order to consistently refer to the results of the previous chapter and of both spatial and temporal parameters. The hypothesis were tested on the laps one to twelve. The final lap differed mostly significantly from them and was separately considered. Hypotheses were accepted when trends could be illustrated with linear or quadratic interpolations and there were significant differences in the lap means of all runners. In order to prove ($H12$), CT was evaluated for each runner by determining the mean of each lap. These mean values were lap-wisely compared between all runners. A linear regression was applied to the average data. FT was analyzed the same way ($H13$), but a quadratic regression was applied to its averaged data. CT and FT was further normalized for SS and then linear regression analysis was applied to it ($H16$, $H17$). The duty factor ($CT \cdot SR$) and the ratio FT/ST were calculated for each runner, too, afterwards their means of each lap were also compared inter-individually ($H14$, $H15$) and then both factors compared in a plot with their linear regressions. The duty factor and the ratio FT/ST were also normalized for SS. The behaviors of the normalized parameters were described (DFn, $FTSTn$).

SLn and SRn were calculated for the averaged data by normalizing SL and SR for SS. Linear regressions were used to show trends within four phases ($H8$). SRI was determined for $H9$ by applying the averaged data to eq. 7.1. It was expected to have values between 0 and 100 % numbering the proportion of SR in the rate of change of SS. RL (SL/SR) was applied on the averaged data ($H10$). Therefore, a quadratic regression described its behavior. For $H11$, the *HR:1/CT* index was calculated by multiplying the means of HR and CT for each runner and lap and then plotted.

$$SLI = SL \frac{\log \frac{SF_{i+1}}{SF_i}}{\log \frac{SS_{i+1}}{SS_i}}, \quad SS \ldots \text{stride speed} \quad i \ldots \text{ith stride in timeseries} \tag{7.1}$$

For $H18$, the optimal stride length (SL_{opt}) was determined by the product of body height (BH) and the factor 1.15 according to Scholich (1978). In order to allow for comparison, other calculations incorporating a proportion of the leg length (LL) were tested, too, i.e., $1.40 \cdot LL$ (Svedenhag & Sjodin, 1994) and $2.11 \cdot LL$ (Hoffman (1965) cited in Elliott & Blanksby (1979)). SL of each runner was normalized for SL_{opt} and then compared for each lap between all runners. Over-striding was defined as SL being greater than SL_{opt}. Optimal SR was assumed to 1.5 Hz according to Daniels (2005). As an alternative, the optimum at 1.35 was tested, too, according to Cavagna *et al.* (1997).

The analysis of variance (ANOVA) was used to test for differences between the laps if single runners were lap-wisely examined. The significance level was set to 0.05. Plots were created for visualization, in order to allow for a qualitative interpretation of the data, too. In order to evaluate the changes in the stride pattern, it is noteworthy to bear in mind the time course of the average running speed—there was a decrease from lap one (3.80 $m\,s^{-1}$) to lap ten (3.40 $m\,s^{-1}$), then a slight increase until lap twelve (3.48 $m\,s^{-1}$) and another increase towards and in the final lap (4.47 $m\,s^{-1}$).

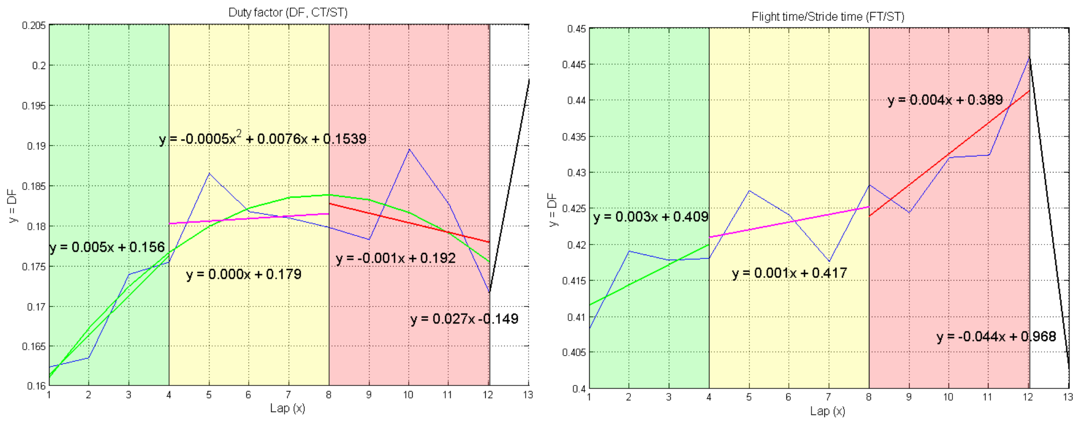


(a) (b)

Figure 7.1: (a) CT (b) FT

7.3 Results

The overall means of CT and FT were $0.1483 \pm 0.0110\,s$, and $0.5819 \pm 0.0223\,s$, respectively. CT increased gradually ($CT = 0.0012x + 0.14$, where x is used to denote the laps in all regressions) until lap eleven, then reached the peak value of $0.18\,s$ in lap twelve, and steeply decreased to approx. $0.13\,s$ during the end spurt, see fig. 7.1 a. CT of all runners with two exceptions (6, 8) exposed significant differences in the means between laps. For the average data and until lap twelve, $H12$ was not accepted due to the two exceptions. FT did not demonstrated a strictly monotonic trend and therefore a quadratic regression was used: $FT = -0.0012x^2 + 0.016x + 0.55$. The maximum of the regression was reached at lap seven. FT had a peak in lap seven and nine, see fig. 7.1 b. Therefore, $H13$ was rejected. Nevertheless, this parameter was significantly different in its lap means for all runners with three exceptions (runner: 6, 8, 9). The overall means CTn and FTn were 0.051 ± 0.0056 and 0.166 ± 0.160, respectively. Both CTn and FTn increased until lap twelve ($CTn = 0.0011x + 0.036$, $FTn = 0.002x + 016$), then both decreased steeply below their start values. Therefore, $H16$ was accepted and $H17$ rejected. The normalized parameters, CTn and FTn, had a similar trend compared to their originals, whereas the quadratic regression fits stronger the time course of the FTn than of the FT (norm residuals: 0.03 and 0.06, respectively).



(a) (b)

Figure 7.2: (a) CT/ST (b) FT/ST

The mean duty factor (CT/ST) was 0.1787 ± 0.0010. The overall trend was an increase from the beginning to the end ($0.017x + 0.17$). There were two local maximums in lap five and eleven respectively. In between them, there was a local minimum near its overall mean. In lap twelve, it decreased to approx. two thirds of its STD and reached its global maximum at 0.1982 in the final lap. Until lap twelve, it could also be interpolated with a quadratic function ($-0.00048x^2 + 0.0076x + 0.15$) and a very low norm residual (0.014) compared to the linear regression (0.023), see fig. 7.2, a. Therefore, $H15$ was rejected. There were significant differences in the lap means of all runners except two runners (8, 9). The mean FT/ST was 0.4229 ± 0.0111. Its linear regression was $FT/ST = 0.0011x + 0.42$ i.e., it increased gradually. In lap thirteen, it decreased below the value of lap one. Therefore, the linear regression without the final lap displayed a stronger increase ($0.0023x + 0.41$). $H15$ was rejected, see fig. 7.2, b. Nevertheless, significant differences were present, except for three runners (6, 8, 9). The ratios CT/ST and FT/ST partly displayed a mutual and diametrical behavior. For example, between lap eleven and twelve DF decreased while FT/ST increased and in the end vice versa. The mean DFn and $FTSTn$ were 0.057 ± 0.004 and 0.1189 ± 0.0108, respectively. DFn revealed a negative U-shaped function ($DFn = -0.00027x^2 + 0.0043 + 0.037$, with a norm residual of 0.005). The maximum (0.054) of the regression was in lap eight. The value of lap one and two were approx. the same (0.042 and 0.044, respectively). $FTSTn$ displayed a linearly increasing trend until lap twelve ($FTSTn = 0.0018x + 0.11$, norm

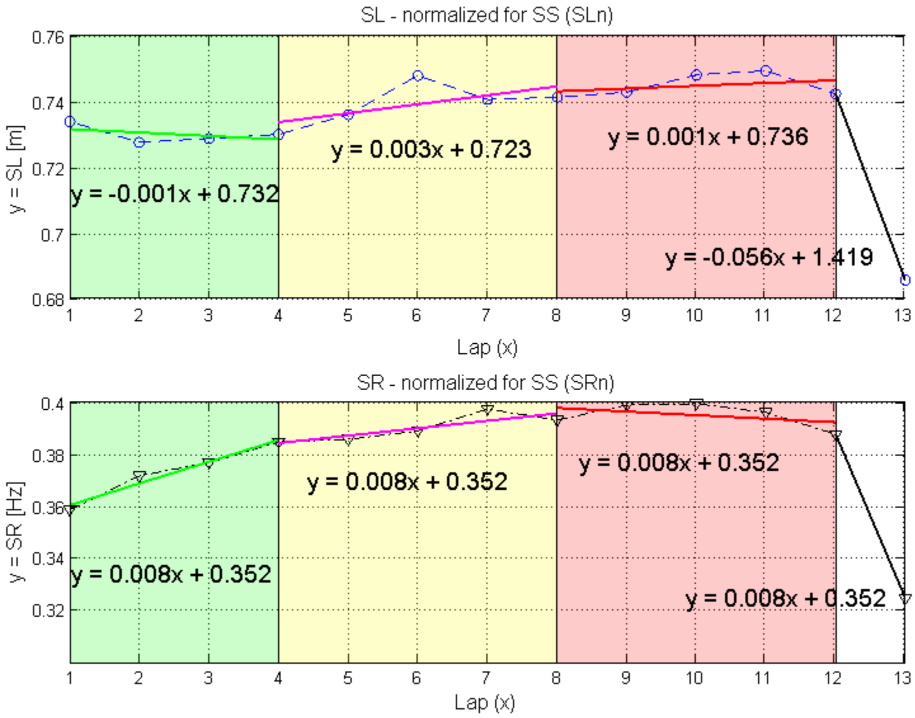
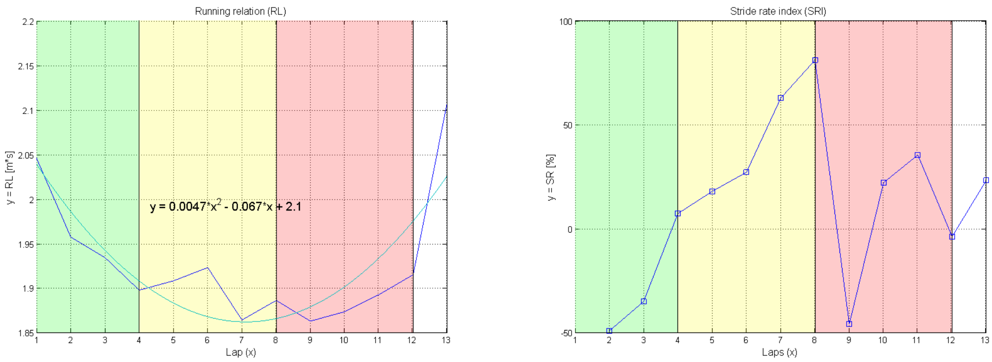


Figure 7.3: *SRn* and *SRn*

of residuals=0.007). The normalized ratio FT/ST had a smoother linear increase than its original.

The normalized parameters SLn and SRn demonstrated diametrical and commensurately trends. In the first four phases, SLn decreased ($SLn = -0.001 \pm 0.732$), while SRn increased ($SRn = 0.008 \pm 0.352$). In phase two (lap four to eight), both of them increased ($SLn = 0.003 \pm 0.723$, $SRn = 0.003 \pm 0.373$), but between laps eight and twelve SLn increased ($SLn = 0.001 \pm 0.736$), while SRn decreased ($SRn = -0.001pm0.408$). The overall trend in both increased until lap twelve ($SLn = 0.0117 \pm 0.73$, $SRn = 0.0028 \pm 0.37$). In the end spurt, both parameters steeply decreased, see fig. 7.3. At all, both parameters differed from their originals, therefore $H8$ was rejected. SRI reached values below 0 before lap three and in lap eight. Between lap three and seven there is an increase in the participation of the SR in SS to approx. 80 %. After the drop of lap eight, it raises to approx. 30 % and 40 % in lap nine and ten, respectively. Before the last lap, it reached about 0 % contribution to changes in speed, but once again in-



(a) (b)

Figure 7.4: (a) *SRI* (b) *RL*

creases to 30 % during the end spurt, see fig. 7.4, a. $H9$ was rejected due to the mutual phases of SL and SR. The overall mean RL was 1.93 ± 0.073. Its quadratic regression equation was $RI = 0.0047x^2 - 0.067x + 2.1$, see fig. 7.4, b. There was no statistical difference between the first and the last lap, but several laps in between statistically differed from each other. The minimum of the regression was in lap seven, the overall minimum of the RL in lap nine. $H10$ was accepted. The overall mean $HR/1{:}CT$ was 1254 ± 71. After reaching a local maximum in lap two, it decreased gradually until lap ten ($HR/1{:}CT = -23x + 1600$) and increased and reached the global maximum in last lap, see fig. 7.5. $H11$ was rejected.

The mean overall over-striding was 1.40 ± 0.08 ($40 \pm 8\,\%$). The first peak of the over-striding was in the lap one. From then on, the amount of over-striding decreased ($SL/SL_{opt} = -0.008x + 1.1$), but was never below 19.7 %. Again, it steeply increased in lap ten and reached its maximum in the final lap, see fig. 7.6, a. If SL_{opt} was set to $1.40 \cdot LL$ and $2.11 \cdot LL$, then the mean ratio were $2.16 \pm 0.13\,\%$ and $1.41 \pm 0.08\,\%$. The mean ratio SR/SR_{opt} was 0.91 ± 0.02. The overall ratio decreased until lap eleven with $-0.0024x + 0.92$. For these laps, the mean was 0.90 ± 0.01. Then increased and reached 98 % of the optimum, see fig. 7.6, b. If SR_{opt} was set to 1.35, the mean ratio was 1.0 ± 0.02. Until lap eleven, the linear regression was $SL/SL_{opt} = -0.0026x + 1$ and the mean ratio was 1.0 ± 0.01 and 1.07 in the final.

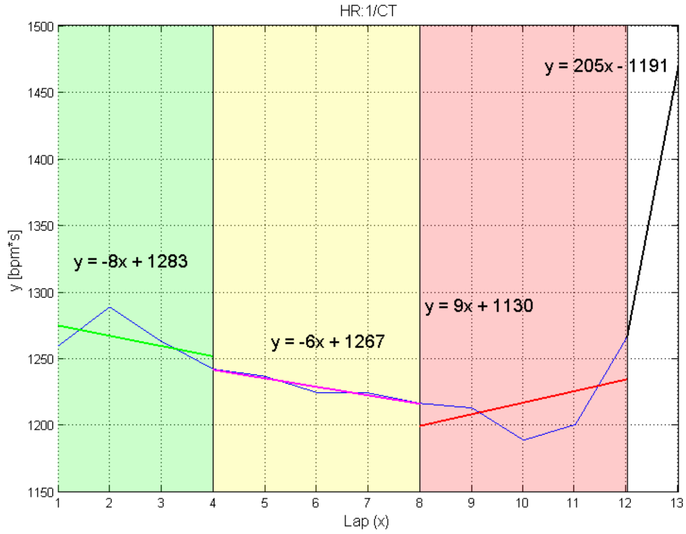
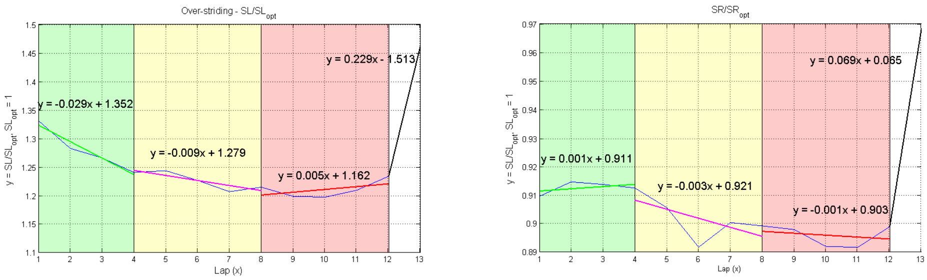


Figure 7.5: *HR:1/CT*

7.4 Discussion

The aim of this study was to extend the knowledge of the previous chapter. The study made use of inertial sensors to obtain temporal stride parameters and a fused measurement system comprising these inertial sensors and a radar device, in order to obtain the spatial parameters. These parameters were validated for their usability in a biomechanical assessment as indicators of fatigue during a full-exhaustive run. Inertial sensors are small and light-weighted, and therefore do not disturb the runner during the training or competition. In general, it is more easy to determine temporal than spatial stride parameters. Though it would be of advantage to realize an accompanying training system for runners with only inertial sensors and suitable temporal parameters. If further spatial parameters would be necessary, then data and the computational effort would increase—however, such a training system to date does not exist.

The results suggested that such a training system could benefit from the integration of CT. As it was expected from the literature review, its increase is a significant indicator of the reduced neuromuscular efficiency. CTn did display the same behavior and may help to assess fatigue as well e.g., when runners are prescribed to run intervals with different paces. The hypotheses were rejected because two runners did not display



(a) (b)

Figure 7.6: (a) SL_{opt} (b) SR_{opt}

significant changes in the stride parameters. Nevertheless, it can be concluded that in the majority of the runners this effect would be apparent. There were also significant changes in the duty factor of the majority of the runners. Its time course appeared to consist of a negative U-shaped part until the finish and a linear raise at the end. Near the end, there was a decrease in SR or an increase in its inverse, ST, and therefore, although CT increased gradually, the duty factor decreased. Its interpretation might be that the reduced neuromuscular efficiency causes higher forces generated by the muscles, in order to stride further—thus adversely affecting the performance and the injury risk. The normalized duty factor with its negative U-shaped regression and the RL with its positive U-shaped interpolation might represent the same kind of adjustments due to fatigue. RL involved a spatial and temporal parameter, whereas DFn involved two temporal parameters and the running speed, which is a derived from a spatial parameter. Chapter 4 concluded that the current realization of the mobile training system of the thesis comprising a customary inertial sensor is not able to precisely determine the stride length but the mean running speed over several strides. The integration of a high-grade sensor inertial sensor engaged with the determination of CT and the mean running velocity from this sensors or combined from a GPS system could be realized in the future. The remaining parameters did display significant changes over the time course in the majority of the runners. For example, FT and FTn displayed a negative U-shaped function, too. The ratio FT/ST increased linearly to the end then decreased linearly. FT/ST and $FTSTn$ increased gradually until the end spurt. This was possible due to the increase in ST and its therefore reduced inverse SR. CT

increased while SS decreased. Therefore, FT could not be used up for efficient forward propulsion. The temporal parameters have a potential to predict the neuromuscular efficiency, whereas such a training system should integrate one of them.

If the interplay between SL and SR is of interest, then their normalized parameters help to reveal the runner's underlying optimization strategy i.e., the changes in the stride pattern. In this study, the runners compensated very early with SR and a very fatigued state through further striding. Further investigations on this stride pattern are warranted, especially in high quality runners. It is guessed that the effectiveness of training could be improved when the stride pattern will be adjusted towards the predictions of the high quality runners. Another view on the same phenomena could be achieved with RL. The decrease in SL mainly caused the stretch of the U-shaped RL time course, and therefore, is suggested to use for the observation of the interplay between SL and SR while focusing on a flat U-shaped time course, in order to keep the overall variability in both stride parameters on a low level. With regard to the minimum energy hypothesis, it is more efficient to keep the stride variability small (or nearly constant) as it was observed in walking. However, running is a fatiguing exercise, and therefore, compensation could be clearly examined in this study.

SRI might also reveal further adjustments of the stride patterns. Herein, it guessed that an optimal stride pattern would display less peaks during the run. This parameter is a very sensitive measure. In running, there are higher speeds compared to walking and under fatigue, there might be also a higher rate of changes in the speed. These higher changes and the variability from the measurement system might have caused these unexpected results of SRI. Nevertheless, it showed that SL is responsible for speed changes at the beginning and near to the end, whereas SR did in between. It could be reasoned that, in phase one, the runners compensated unconsciously with an increase in SR, while they attempted to keep to a high running speed by high values of SL, and therefore, obviously reacted with a high SL to a decrease in speed. In the middle of the time-trial, both parameters decreased in their absolute values, even though the SR was responsible for the rate of changes in speed. In the last laps, SL dominated, which is thought to be accompanied with high volition of the runners. Herein, the SLI or SRI index may enrich the biomechanical assessment.

No matter how the over-striding was calculated, the study revealed that its recreational runners did over-striding at a high level—even under high fatigue when performance and the SL was drastically decreased, there was at least 20 % (or even about 35 %) over-striding. This running technique might come together with lower than optimal SR. Under fatigue, the difference from the optimal grew and at highest effort (in the final lap) the optimum could neither be reached for SL nor for SR. Such a running technique is known to reduce performance and to increase the injury risk. SL with regard to its optimal determined by anthropometric measures might be an important tool in the training. Herein, the determination of SL is compounded, whereas SR can be more easily obtained. Taken collectively, a training system revealing these two indicators can help to analyze the running technique and thus supports to select training methods for improving the stride pattern. For example, to the author's knowledge it is also imaginable to observe the stride pattern during the training and to force the runner in a more preferable stride pattern (e.g., smaller SL and higher SR) and to break up when the runner can no more keep to this pattern, in order to gradually accustom to an optimal pattern.

In a full-exhaustive time-trial, *HR:1/CT* index does not provide new information than others have already given. This index might be useful during runs of moderate to high intensities. Nevertheless, the *HR:1/CT* index was expected to gradually increase, because the effort of the runners was permanently high. All the runners forced themselves to full-exhaustion during the time-trial. However, the time course of the stride parameters and the variability in all parameters might have given recovery from fatigue. At least the *HR:1/CT* index suggested that the runners could recover when comparing the first two laps until lap ten. Obviously, the recreational runners started too fast and therefore had to slow down what in fact ensured recovery. From this perspective, this index also demonstrated that the runners could not keep to a constant effort level and that they were only able to raise their effort to a higher level during the last three laps. Further investigations could concentrate on exact values for this parameter. If a scale could be created for a typical middle-distance run, then there would be a further tool, which could be used for advising the runner during high fatiguing exercises.

7.5 Conclusion

It could be shown that a biomechanical assessment based on the analysis of the stride pattern allows for drawing conclusions upon running technique and fatigue and moreover, to give suggestions for the enhancement of performance and for reducing the injury risk. From the data, it became apparent that the stride pattern in a $5\,km$ full-exhaustive run can be predicted. A training system that aims to support runners during training towards the latter suggestions may take benefits from incorporating these parameters.

8

Stability of stride kinematics in a 5 km time-trial

8.1 Introduction

From the last two chapters, it became apparent that analyzing the time course of the stride parameters may support the assessment of the performance and the injury risk under the influence of fatigue during a full-exhaustive run. The time-trails were divided into three even main and a final phase. Then the analysis of the stride parameters were applied to these (four) phases and compared. The analysis of the variability with measures of its degree supported the estimation of the effects of fatigue and the conclusion upon the motor potential. Although the mean of the degree of variability did not change remarkably, trends in the time course and in its variability could be observed, which were analyzed with regressions. The assessment of the adjustments in the stride parameters allowed for evaluating the running form and estimating the effects of fatigue on the runners. While the heart rate was on a stable level between at least 3 % above the anaerobic threshold and close to its maximum, there were significance biomechanical adjustments such as the decrease of the stride length (SL) until the middle of the run, though over-striding and a decrease of the stride rate (SR). Apparently, a biomechanical assessment would enrich the assessment of runners during training or competition. At first glance, fatigue was considered as the impairment to produce the desired force to generate forward propulsion. Those effects might have emerged with the depletion of energy depots (e.g., glycogen in the muscles) but also with a reducing neurotransmitter activity. The adjustments observed were mainly associated with short term effects on performance. The purpose of this study was to go further while analyzing the stability of the movements. It was the intention to provide the biomechanical assessment of runners with measures with respect to the neuromuscular system. The purpose was to describe the effects of a severe insensitive middle-distance run on a measure of stability.

Training is essential for athletes. Its overall goal is to meet the level of the stimuli, which allows for accommodation to training load, in order to enhance performance. Training and recovery have to alternate in a still not fully understood manner, in order to reach the training goal. The accommodations in the several underlying systems involved in the training rely on different recovery periods of time. Therefore, the training must be chosen carefully, when its schedule is aimed to be maxed out. After an intensive training load, it takes only few days and the energy depots of the muscles have refilled, however, it takes longer to recover for the passive structures of the movement apparatus such as tendons while the nervous system depends on the longest recovery period (Bishop *et al.*, 2008). In this context, especially elite runners can experience the phenomenon of overreaching and overtraining (Lehmann *et al.*, 1991). Both of them foster over-use injuries. The incidence rate of injuries in recreational runners does not remarkable differ from the elite runners. Therefore, inappropriate training (i.e., too intensive and/or too frequent) and poor running technique have been mainly found responsible for the phenomenon. Insufficient recovery periods might disturb the efficiency of the neuromuscular system and thus the stability of the movements. Meardon *et al.* (2011) demonstrated that the movements of runners with a history of injuries were less stable compared to runners without a history of injuries. Regarding the statistics of over-use injuries (Anderson, 1996), the majority of runners will suffer from an injury within one year. Measurements of the stability may indicate when the motor potential of runners is reduced and, therefore, the risk of injury might have increased.

Stable movements are characterized by a certain amount of variability, whereby the structure of the variability has a key function for the stability. A movement is considered stable when it can be adjusted to sudden changes in the environment or in the propelling system (higher work load) (Dingwell *et al.*, 2010). The fluctuations of strides, especially in the gait of walking were found to be connected in a nonlinear and dynamic manner i.e., strides displayed long-range correlations. The correlations were weaker when subjects were forced to walk under metronome conditions (Terrier *et al.*, 2005), were placed to high cognitive load (Beauchet *et al.*, 2005) or suffered from lower extremity injuries (Heiderscheit *et al.*, 2002). In the literature of treadmill running, several estimations of stability were applied to the stride time (ST) time series. The recreational runners in the only existing over-ground running study of Meardon *et al.* (2011) obviously reduced their stability (tested on the ST) during a middle-distance run at high intensity. Therefore, the aim was to validate their results, in order to allow for generalization. The measurement methods *IMU/GPS* were applied to extend this study while conducting also an analysis of the remaining stride parameters. A subordinated goal was to compare the stability calculated with several stride parameters. The

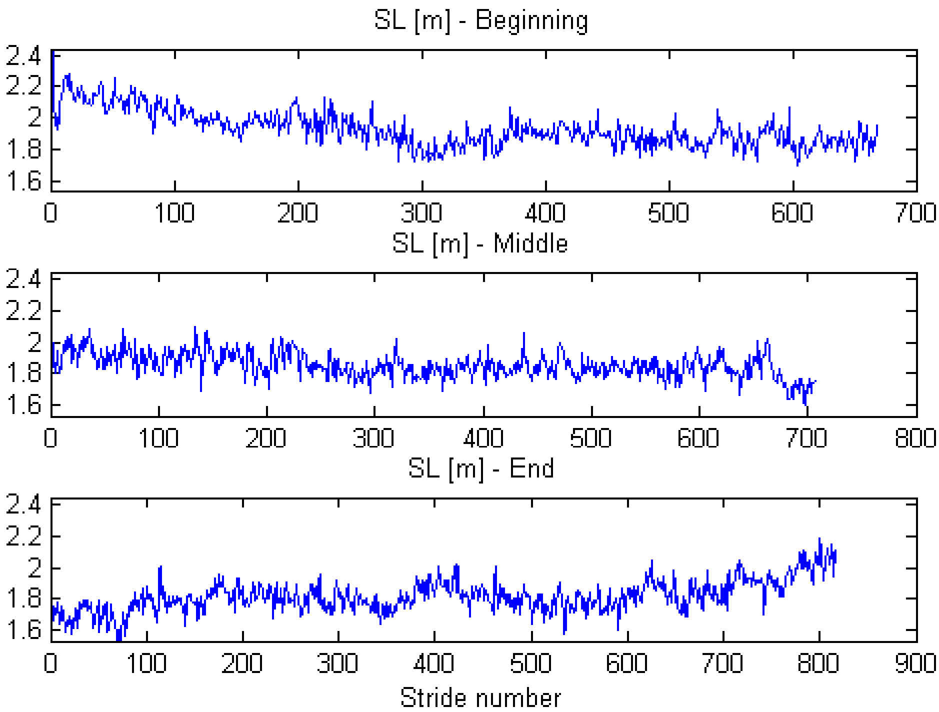


Figure 8.1: Raw data of *SL* from *IMU/GPS* within three phases. *DFA* was applied to each phase.

dependency of the spatial and temporal parameters may suggest that both of them would allow to draw the same conclusion on the stability of the strides. If this could be confirmed then, for example, a computer-aided training system could make use of only temporal parameters, thereby reducing computational effort, data transfer and storage and associated requirements. Strictly speaking, the purpose was to investigate to which extent long-range correlations can be found in the typical stride parameters. Additionally, the effect of fatigue on the strength of the long-range correlations was aimed to scrutinize. The ultimate goal then was to relate the amount of variability and the measure of stability.

8.2 Method

The analysis was conducted on the results of the 5 km time-trails. The run of subject 9 had to be excluded due to a partly data loss. Therefore, the study involved ten runs of recreational runners. The detrended fluctuation analysis (*DFA*) was chosen to estimate the stability of the stride parameters. *DFA* is a robust measure for nonstationary time series (including signals with time dependency) and allows as a mono-fractal method for the estimation of long-range correlations of fluctuations in time series (Terrier & Deriaz, 2011). *DFA* is an extended root-mean square calculated over sliding boxes with increasing size—starting with 10 samples and increasing by 10 samples until the number of samples is reached. Thereby, in each window the trend was removed by subtracting its mean values yielding F_n. The natural logarithm of the integrated time series y_n (over these windows with removed mean) was plotted against the natural logarithm of F_n. If the slope of this plot is linear, then it equals the coefficient α. Ant-persistent correlation is defined by values of α smaller than the border of white noise of 0.5, greater and before unity it is positive correlated, whereas above unity until 1.5 the correlation fades into Brownian noise. It was hypothesized that the fluctuations of the stride parameters exhibit a long-range, fractal-like correlation pattern under fatiguing running conditions ($H23$). These correlations were further hypothesized to decrease when individuals fatigue near to full-exhaustion after the middle of the run i.e., the fluctuation pattern becomes random ($H24$). The algorithm is presented below (adapted from Chau (2001a, p.62)):

1. Integrate the time series by computing $y(k) = \sum_{i=1}^{k} x_i - \overline{x}$
   where x is the sample mean and $k = 1, \dots, N$.

2. Divide $y(k)$ into intervals of equal length n.

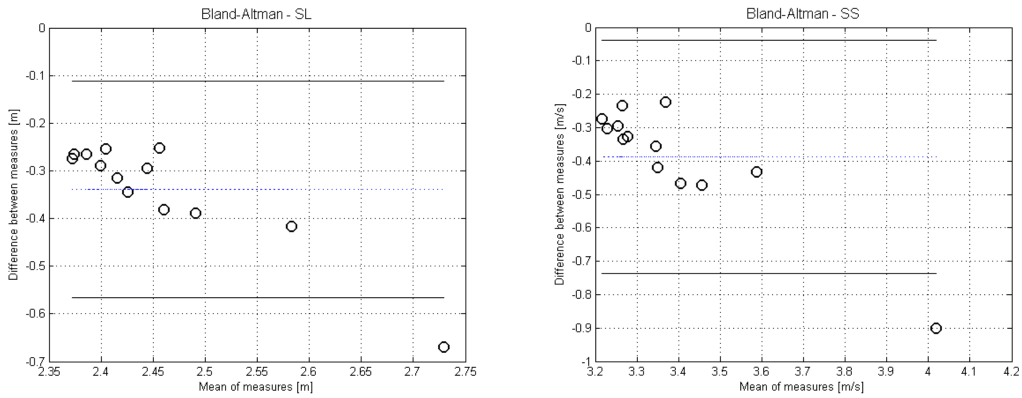
3. Fit a least square line to the data in each interval. Let $\overline{y}(k)$ represent the y-coordinate of this line.

4. Compute the average fluctuation of $y(k)$ about the locally best fit line.

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}(y(k) - \overline{y}(k))^2} \qquad \overline{y}(k) \dots \text{mean of y(k)} \tag{8.1}$$

5. Repeat steps 2 through 4 for different values of n.

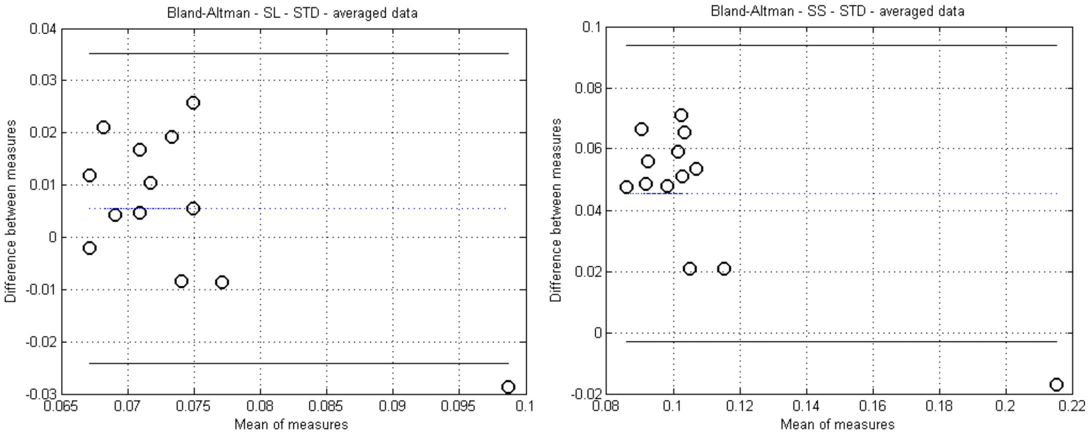
6. Plot $\log F(n)$ versus $\log n$. If a linear relationship is evident, compute the slope α of the line.



(a) SL (b) SS

Figure 8.2: Bland-Altman plot to compare SL from *IMU/GPS* and *IMU/RADAR*. Both SL and SS display an off-set to the reference measurement system. As the runner increases speed or strides further, the offset increases, too. Except for one outlier (occurred during end spurt) in each plot, all data points are within the 95 % limit of agreement.

ST was determined between to consecutive stance phases. The contact time (CT) equaled the stance phase. The flight time (FT) was the difference of ST and CT. The parameters SL and stride speed (SS) were considered carefully, because the speeds of the recreational runners were in the sensitive speed range of the measurement method. Therefore, the measurement results of the *IMU/GPS* were compared to the results of the *IMU/RADAR* system, see chap. 6, with the means, the standard deviation (STD) and the coefficient of variation (CV) of the differences of the SL and the SS. Furthermore, Bland-Altman plots were evaluated. The data obtained by the *IMU/GPS* system was present for the left and the right foot. The inner foot i.e., the left one, was selected for comparing the runners and the laps. The run was divided in three phases (beginning was from lap one to four, middle was from lap five to eight and the end phase was from lap nine to thirteen). Descriptive statistics, coefficient of variation (CV) were applied and compared to the results of the study of Meardon *et al.* (2011). A one-way repeated measures analysis of variance (ANOVA) was used to test for differences between the phases of the run (beginning, middle, end). As a physiologic measure, the heart rate (HR) of each runner was normalized to the 3 % above anaerobic threshold (HR_{T2n}) and then the mean and the STD were calculated for each phase and for the overall



(a) *STD* of *SL* (b) *STD* of *SS*

Figure 8.3: Bland-Altman plot to compare *STD* of *SL* and *SS* from *IMU/GPS* and *IMU/RADAR*. Except for one outlier (occurred during end spurt) in each plot, all data points are within the 95 % limit of agreement.

data.

8.3 Results

The overall mean strides and *HR* were 1803 ± 282 and $178 \pm 9\,bpm$, respectively. The mean *HR* in the beginning, middle and end was 171 ± 10, $180 \pm 8\,bpm$, and $182 \pm 9\,bpm$, respectively. The overall mean HR_{T2n} was 1.08 ± 0.06, and in the beginning, middle and end it was 1.03 ± 0.06, 1.09 ± 0.07, 1.10 ± 0.07, respectively, thus *HR* of all runners were permanently 3 % above the anaerobic threshold. The mean differences between the measurement methods *IMU/GPS* and *IMU/RADAR* for *SL* and *SS* were $-0.34 \pm 0.11\,m$ ($CV = -33\,\%$) and $-0.39 \pm 0.17\,m^{-s}$ ($CV = 45\,\%$), respectively. The *IMU/GPS* had a systematic error, which underestimated the spatio parameters. The mean *STD* of *SL* and *SS* were $0.0055 \pm 0.114\,m$ ($CV = 0.002\,\%$) and 0.045 ± 0.024 ($CV = 53\,\%$), respectively, see fig. 8.2 and 8.3. The mean, *STD* and *CV* observed in *ST* were similar to the results of the comparative study, where the overall mean was $0.70\,s$ ranging from $0.67\,s$ to $0.73\,s$ in the non-injured group. *STD* was $0.012\,s$ ranging from $0.007\,s$ to $0.017\,s$. *CV* was 1.7 % and ranged from 1.0 % to 2.5 %. The mean *ST* of each phase during the run of this study is reported in tab. 8.4 and 8.5.

Table 8.1: Effect of a severe intensive run on long-range correlations expressed by the coefficient α determined by *DFA*

Phase	ST	SL	CT	FT	SS
1	0.82 ± 0.17	0.81 ± 0.14	0.78 ± 0.05	0.76 ± 0.10	0.82 ± 0.17
2	0.86 ± 0.17	0.82 ± 0.19	0.82 ± 0.08	0.79 ± 0.12	0.85 ± 0.20
3	0.86 ± 0.16	0.82 ± 0.18	0.81 ± 0.08	0.78 ± 0.14	0.84 ± 0.20
overall	0.86 ± 0.16	0.82 ± 0.19	0.81 ± 0.08	0.79 ± 0.14	0.85 ± 0.21

Table 8.2: Effect of severe intensive run on variability expressed by mean STD (CV))

Phase	$ST\,[s]$	$SL\,[m]$	$CT\,[s]$
1	0.0162 ± 0.0072 (7.7 %)	0.0952 ± 0.0194 (8.2 %)	0.0105 ± 0.0028 (17.5 %)
2	0.0152 ± 0.0044 (6.6 %)	0.0850 ± 0.0268 (9.0 %)	0.0103 ± 0.0022 (16.2 %)
3	0.0154 ± 0.0087 (6.6 %)	0.0732 ± 0.0380 (11.5 %)	0.0100 ± 0.0049 (12.9 %)
overall	0.0156 ± 0.0067 (6.9 %)	$0.0844 \pm 0.0028\pm$	0.102 ± 0.003 (15.5)

The coefficient α of the stride parameters and phases is reported in tab. 8.1. The mean α of all stride parameter was 0.83 ± 0.03. However, they did neither remarkably differ in the overall data nor between the phases. For α, none of the stride parameters significantly differed between the phases ($p = 0.82 \pm 0.17$) but exposed strong long-range correlations ($\alpha \approx 0.8$), see fig. 8.4, a. This is in contrast to the results of Meardon *et al.* (2011), see fig. 8.4, b. Both groups, the injured and the non-injured recreational runners, showed a downwards trend of α. The mean STDs were in the same range as the fused radar and IMU measurement method revealed (about 10 %) but were greater than in the comparative study, see tab. 8.2 and 8.2. SL is illustrated as an example of the raw data in fig. 8.1.

8.4 Discussion

The ultimate goal of this study was to examine the effect of a full-exhaustive run on the strength of long-range correlations of the stride parameters. Little is known in the research of the stability in over-ground running, especially spatio parameters have not yet been investigated. This study could obtain the spatio parameters from over-ground running. The engaged measurement system was an extended pedestrain dead reckoning approach (PDR), which was extended to the requirements of running. The temporal parameter ST was similar to the comparative study. Nevertheless, the requirements

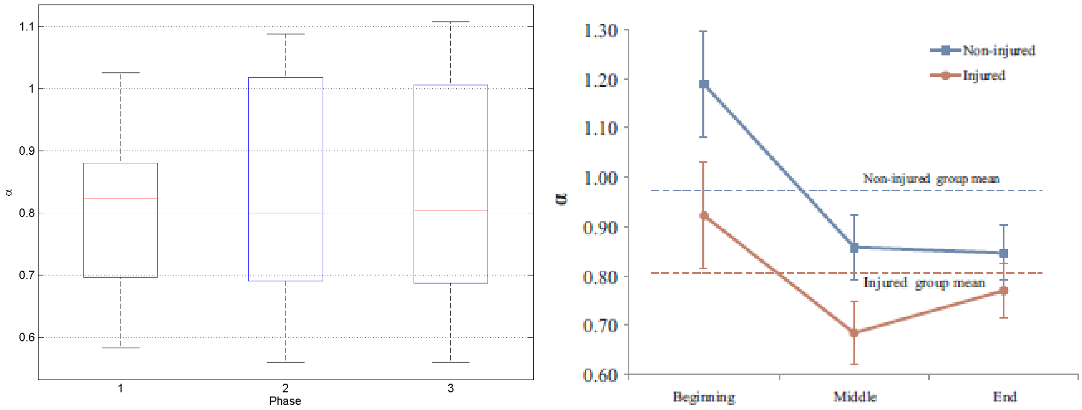
Table 8.3: Effect of severe intensive run on variability expressed by mean STD (CV))

Phase	$FT\,[s]$	$SS\,[m^{-s}]$
1	0.0156 ± 0.0052 (9.2 %)	0.1637 ± 0.0573 (11.9 %)
2	0.0143 ± 0.0058 (8.7 %)	0.1478 ± 0.0487 (11.6 %)
3	0.0121 ± 0.0073 (8.9 %)	0.1396 ± 0.0680 (13.8 %)
overall	0.014 ± 0.0060 (8.9 %)	0.15 ± 0.058 (12.4)

of the speeds in the middle-distance running were challenging for this method to determine the spatial stride parameters. The absolute errors of the *IMU/GPS* system were remarkably but systematic i.e., the trends and the variability observed with the comparative measurement system were similar for SL and SS. As it was expected from chap. 5, the error increased with further striding and speed. The analysis method used for the observance of the mono-fractal like behavior in the time series removed the trend in the data. Therefore, this absolute error might have been reduced. CT was only a fragment of the stance time, therefore, it was smaller than in other studies but also demonstrated remarkable variability and thus FT was higher than in other studies.

The effect of fatigue on stride parameters were lower in this analysis than illustrated with descriptive statistics and regression analysis (see chap. 6 and 7). Here, the run was divided into three even parts, whereby the last part included also the end spurt. Due to the averaging of the continuous data within these parts, the trends were diminished. However, the purpose was to apply *DFA* to the individual runners and then compare the results to the analysis of variability. In contrast to the comparative study, there were no changes in the coefficients of stability. It is of note, that in all parameters and in the majority of the runners the coefficients illustrated a strong correlation. If it is assumed that *DFA* was capable to determine the stability in the stride parameters, then the conclusion is that most of the runners did demonstrate stable movements over the whole run. All runners took part in the time-trail as they were rested. The comparative study did not tell details about the resting of the runners in before the run.

Meardon *et al.* (2011) stated that the values of α in their study were higher as in treadmill running (Jordan *et al.*, 2007a, 2006). The movements in treadmill running might be very stable, because the treadmill induces them. Thereby, less cognitive



(a) (b)

Figure 8.4: DFA was applied to (a) the time series of SL of the 5 km time-trail and to ST of a run to full exhaustion at 5 km pace in the study of Meardon *et al.* (2011) The results of the other stride parameters were similar (a). In contrast to (b), there was no trend in α but the range spread out.

load seems to be necessary. The issue with DFA is the comparability between runners and between trails. For example, in the study of Meardon *et al.* (2011), the runners approached to Brownian noise in the first phase i.e., demonstrated reduced stability in beginning. In the middle of the run, they reached the stable zone but were above the optimum of 0.8. It could be reasoned that the runners forced their movements under high volition in the beginning of the race. This might have caused inappropriate movements. Under fatigue, they had to give up forcing and an automatic movement pattern finally dominated. In this meaning, fatigue might also force the runners to run in a natural movement, probably, at least at an early stage of fatigue, and, therefore, might have helped to adjust to a stable movement pattern. Noteworthy is that those runners were attempted to keep to the their individual 5 km pace with a running watch compromising a footpod. Similar to the metronome experiments in walking, the adjustment towards a target stride pattern can even result in anti-persistent correlation. In this study, the runners were not prescribed to run at any pace but only instructed to reach their individual best time. Perhaps, their compensation observed in the variability of their stride parameters could ensure a stable movement. In the middle and end, the runners of the comparative study might have customized to the speed and, therefore, reached nearly stable movements, too. Another issue with stability measures is that at

Table 8.4: Effect of severe intensive run on the stride parameters

Phase	$ST\,[s]$	$SL\,[m]$	$CT\,[s]$
1	0.721 ± 0.055 (7.7 %)	2.28 ± 0.19 (8.2 %)	0.150 ± 0.026 (17.5 %)
2	0.730 ± 0.048 (6.6 %)	2.26 ± 0.20 (9.0 %)	0.153 ± 0.025 (16.2 %)
3	0.727 ± 0.048 (6.6 %)	2.29 ± 0.26 (11.5 %)	0.153 ± 0.020 (12.9 %)
overall	0.72 ± 0.05 (7.0 %)	2.28 ± 0.22 (9.6 %)	0.151 ± 0.024 (15.5 %)

the same time more sources might have an influence on the stability of the movements and it seemed to the author as if it cannot be distinguished whether or not an influence releases load from the neuromuscular control system and thus stabilizes the movements or puts more load on it and destabilizes the movements. In both cases a reduced stability would be indicated. Therefore, it appears as if such a stability coefficient is not a sufficient tool alone but might be a necessary tool in combination with other analysis methods, in order to estimate the runner's movement stability. The neuromuscular control system can partly be reflected with such a coefficient. If the several sources of influences can be determined and evaluated, then obviously such a coefficient gains more value.

It can be further speculated that in the middle-distance running, stability will not be reduced over the time-trail. Further studies are warranted to investigate the effects of a multiple prolonged time-trails with minimal resting e.g., three runs within a week. Due to the slow recovery of the nervous system, it is guessed that then a coefficient of stability is supposed to decrease. During training, a diagnosis of variability and a thereon based intervention aiming at stabilizing the stride pattern seem to be powerful tools, which might support runners. Time series of SL, SR and CT can give incentive hints of adjustments during the run. Compared to better runners, the variability was relatively high. The application of *DFA* to the temporal and spatial parameters yielded similar coefficients near the optimum. From this perspective, a low variability seems to be of advantage, because it reduces energy costs, but does not necessarily disturb the stability of the movement. There were no changes in α during the race. Therefore, it can not be concluded that all parameters are able to depict the overall stability. The stride parameters are highly interconnected, thus, it might be considerable that each of the stride parameters could be enough to estimate the overall stability. Nevertheless,

Table 8.5: Effect of a severe intensive run on the stride parameters

Phase	$FT\,[s]$	$SS\,[m^{-s}]$
1	0.572 ± 0.053 (9.2 %)	3.19 ± 0.3823 (11.9 %)
2	0.577 ± 0.050 (8.7 %)	3.11 ± 0.3599 (11.5 %)
3	0.573 ± 0.051 (8.9 %)	3.17 ± 0.4394 (13.8 %)
overall	0.574 ± 0.051 (8.9 %)	$3.16\pm$ (12.5 %)

also here further studies are prompted, in order to investigate on this topic.

8.5 Conclusion

From this study, it appeared that fatigue has no effect on the long-range correlations of several stride parameters during a full-exhaustive middle-distance run. The runners were recreational athletes and rested. Moreover, the stability expressed by α was at the optimum over the whole time course in the majority of the runners. All stride parameter displayed the same result on stability. It was speculated that only one parameter might be sufficient, in order to determine the overall stability of strides. This is warranted to further investigations. High kinematic variability in all stride parameter does not necessarily disturb the overall stability. Analysis of movement stability such as *DFA* might bring benefits when different trails of one runner are examined, in order to assess the load on the neuromuscular control system. The analysis of the stride parameter and their variability is considered as powerful tool to support the runner during training.

9

Summary and Conclusion

The thesis sought to contribute to the establishment of a biomechanically grounded Mobile Coaching (MC) system for runners. The overall goal was to glean the potentiality and limitations of such a system, in order to allow for a biomechanical assessment of the runners with the to date technical state of the art. The two main goals in sports biomechanics were addressed—increasing performance and reducing the injury risk. The progress in the development of the communication and information technology lent credence to a realization of a computer-aided training system, which should allow to observe a training live and on the Internet and, furthermore, to provide the runners with feedback messages. A prototype of such an MC system has been developed and tested. A variety of low-grade sensors of a commercial Body Sensor Network could be put into use. The version of the MC system for runners incorporated sensors for the estimation of stride derived parameters, the heart rate and the geographic position. The average speed (over several strides), distance and stride rate can be obtained. At this stage this MC system for runners offers a technical solution for the documentation of the training and basis tool for the training analysis. The broad literature review on running analysis revealed that a biomechanical assessment of the running movement would benefit from the analysis of the strides. Stride parameters were found to be important to characterize runners. There was no exclusive set of biomechanical or physiological parameters, which were responsible for influencing the performance or the injury risk. However, in order to enhance performance in sub-maximal speeds the running economy has to be increased. Strictly speaking, from the biomechanically perspective the efficiency must be improved through running technique. An efficient running technique depends also on an optimal kinematics such as stride length and stride rate. Furthermore, optimal adjustments in these parameters would decrease the risk of injury. From other investigators it became apparent that adjustments in the kinematics occur due to or under fatigue, respectively. These adjustments were not consistent across the studies engaged in treadmill, over-ground and different distances. In the middle-distance, only the study of Hanley & Smith (2009) examined the stride kinematics at three time points in a 5 km over-ground run. A model according to Saziorski *et al.* (1987) uttered to predict the time course of the stride parameter length and rate of elite runners in middle-distance time-trail. Although the analysis of variability has gained to provide useful information for economy, the model according to Saziorski *et al.* (1987) did not provide information about the variability of the stride parameters but only that the running speed should be in a range of a few percentage. The analysis of variability had not yet been extensively investigated in over-ground running. While it is aimed to reduce it in running, it also holds functionality for adaptation to any changes around or in the runner. In walking and treadmill running, a variety of studies pointed to the analysis of consecutive strides enabling the estimation of the stride stability i.e., to which extend a runner can adjust its movements towards perturbations. A complex interplay between variability and stability could be seen in these studies on the example of the stride time mostly under metronome (like) conditions or with preferred or slower or faster than preferred gait speeds. In these studies, nonlinear approaches have been used to determine measures of the stride stability. Only the study of Meardon *et al.* (2011) applied such a measure in an intensive over-ground middle-distance run and illustrated and showed that the measure was reducing with respect to time.

The integration of high-grade (analog) sensors in the MC system was foreseen with a micro controller but was not accomplished. However, in order to allow for analysis of consecutive strides in middle-distance running an pedestrian dead reckoning approach was extended. This method (*IMU/GPS*) is promising in the biomechanical assessment of running. Its implementation relied on the stance phase detection. Other solution were tested, too, but not evaluated. For example, to improve the contact time, the initial contact and the toe-off event can be integrated. In order to improve the precision of the system, further conditions might be considered. For example, the weight of the GPS can be increased or that of the projection into the lane. The *IMU/GPS* can also determine the step parameters and support further investigations, which are engaged in symmetry. Acceleration data was present, too, but not evaluated for peaks at initial contact and toe-off. The analysis of these peaks could also enrich a biomechanical assessment. Nonetheless, the analysis of running is challenging due its relative high speeds. The amount of data processed can not be transferred on the Internet within this MC system. These calculation would have to be done on the smart-phone—requiring sufficient power. The detection of the temporal parameters such as the stride rate are less challenging. In order to reduce load on the smart-phone and transfer, one of the goals was to determine whether or not temporal parameter convey the same information on stability than the spatial parameters. From the results, there was no difference i.e, both spatial and temporal parameters indicated stability in the same way. At least the mono-fractal analysis revealed no changes in the stability of the stride parameters. From this study, it cannot be reasoned that both spatial and temporal would indicate a change in stability with the same amount in the stability indexes. It remains an open issue whether or not the several stride parameters can represent special domains of stability.

From the conducted study, it was found that the stride length was responsible for the decrease in running performance. The stride rate compensated but earlier as predicted from the model according to Saziorski *et al.* (1987). A suggestion of kinematic markers was made. These markers displayed critical phases on performance and the injury risk during the run. The recreational runners of this study did over-striding and ran at too low stride rates. Some researchers assume that (recreational) runners do automatically match their optimums through training. Nevertheless, a biomechanical assessment might be helpful for the recognition and to intervene. Under fatigue, the stability illustrated with the detrended fluctuation analysis (*DFA*) did not change in the recreational runners. In the biomechanical assessment, it was regarded as an important factor to estimate the neuromuscular potential within the training cycle. This is should be proved.

Advising runners during running is a critical topic, because different opinions were found in the literature. On one hand side, it was argued that any interfering would decrease the economy and on the other hand side, that runner do not match their optimal stride pattern and rely on the retraining. Running is mainly an unconscious movement, therefore, only little feedback can be suggested to give. From the literature review, it stems that at first the runner relies on a stable and sufficient high stride rate. The appropriate stride length and the sustaining to it comes as a second goal. For these training goals, an MC system should keep the runner to an optimal stride rate and further prevent from over-striding. An increasing contact time can be seen as a simple marker of the effects of fatigue on the reduced neuromuscular efficiency.

The thesis concluded that the analysis of consecutive stride parameters enriches a biomechanical assessment of runners in middle-distance running. It is possible with the adjustments in the stride pattern to draw conclusions upon performance and the risk of injury. Computer-aided training systems were considered as a helpful diagnosis tool to base further suggestions for feedback and training advice on it.

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Appendix A

Declaration

Declaration

I herewith declare that I have produced this paper without the prohibited assistance of third parties and without making use of aids other than those specified; notions taken over directly or indirectly from other sources have been identified as such. This paper has not previously been presented in identical or similar form to any other German or foreign examination board.

The thesis work was conducted from March 2009 to April 2014 under the supervision of Prof. Dr. Arnold Baca.

Vienna, April 4, 2014

Appendix B

Resume

Resume

Personal details

Name:	Sebastian Bichler
Nationality :	German
Place of birth:	Leisnig, Saxony, Germany

Education

2008	Master of Arts in Sport and computer science (M.A.)
2003–2008	University of Technology in Chemnitz, Saxony (sports and computer science)
1992–1999	Grammar school in Doebeln, Saxony
1987–1992	Secondary school in Rosswein, Saxony

Employment record

2013–	Contract researcher at the department of biomechanics and physiology of the University of Stuttgart
2009–2013	Research assistant at the department of biomechnics, kinesiology and applied computer science of University of Vienna
2007–2008	Software engineer at Logicpd
2006–2008	Student assistant at the department of sport medicine of the institute of sport science of the Technical University of Chemnitz.
2003–2007	Student assistant at the Institute of Mechatronic e.V., Chemnitz, Saxony

Appendix C

IMU/RADAR measurements

Table C.1: *SS*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	3.4	3.5	-	3.8	3.8	3.8	3.5	3.4	4.7	3.8	4.3	3.8	0.4
2	3.1	3.1	4.7	3.8	3.7	3.7	3.4	2.8	4.5	3.2	4.4	3.7	0.6
3	3.1	3.0	4.3	3.7	3.7	3.7	3.4	2.7	4.5	3.4	4.4	3.6	0.6
4	3.1	2.8	4.2	3.6	3.8	3.6	3.4	2.6	4.4	3.5	4.2	3.6	0.6
5	3.0	2.9	4.1	3.6	-	3.6	3.4	2.6	4.3	3.6	4.1	3.5	0.6
6	3.0	2.9	-	3.5	3.8	3.6	3.4	2.5	4.2	3.3	4.3	3.4	0.6
7	3.1	2.8	-	3.5	-	3.6	3.4	2.5	4.1	3.3	4.2	3.4	0.6
8	3.0	2.8	4	3.4	3.7	3.5	3.3	2.4	4.0	3.4	4.3	3.4	0.6
9	3.1	2.6	3.8	3.6	3.7	3.8	3.3	2.4	-	3.5	4.1	3.4	0.5
10	2.9	2.8	3.6	3.3	3.5	3.6	3.4	2.3	4.1	-	4.1	3.4	0.6
11	3.0	2.8	3.5	3.4	3.5	3.7	3.5	2.5	3.9	3.4	3.9	3.4	0.4
12	3.2	2.8	3.9	3.6	3.6	3.6	3.5	2.4	4	3.7	4.1	3.5	0.5
13	3.3	3.3	5.0	5.3	5.3	4.2	5.1	3.0	-	-	5.6	4.5	1.0
Mean	3.1	2.9	4.1	3.7	3.8	3.7	3.5	2.6	4.3	3.5	4.3	3.6	0.6
STD	0.1	0.2	0.5	0.5	0.5	0.2	0.5	0.3	0.3	0.2	0.4	0.3	

Table C.2: *STD* of *SS*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	0.06	0.06	-	0.09	0.08	0.12	0.08	0.08	0.14	0.27	0.07	0.10	0.06
2	0.08	0.07	0.07	0.07	0.07	0.07	0.04	0.10	0.05	0.06	0.07	0.07	0.01
3	0.09	0.06	0.12	0.04	0.09	0.09	0.06	0.10	0.05	0.07	0.06	0.07	0.02
4	0.06	0.05	0.12	0.05	0.08	0.06	0.07	0.09	0.06	0.13	0.07	0.08	0.03
5	0.06	0.05	0.06	0.07	-	0.05	0.06	0.09	0.05	0.10	0.08	0.07	0.02
6	0.06	0.10	-	0.04	0.06	0.08	0.08	0.06	0.06	0.06	0.05	0.06	0.02
7	0.07	0.05	-	0.05	-	0.06	0.06	0.06	0.03	0.09	0.09	0.06	0.02
8	0.12	0.07	0.06	0.11	0.06	0.05	0.06	0.11	0.17	0.16	0.08	0.09	0.04
9	0.03	0.09	0.04	0.05	0.03	0.07	0.05	0.05	-	0.08	0.07	0.06	0.02
10	0.09	0.08	0.06	0.06	0.08	0.05	0.05	0.10	0.06	-	0.08	0.07	0.02
11	0.07	0.09	0.04	0.05	0.08	0.06	0.09	0.08	0.04	0.10	0.08	0.07	0.02
12	0.05	0.08	0.04	0.12	0.08	0.04	0.15	0.07	0.07	0.09	0.09	0.08	0.03
13	0.11	0.07	0.28	0.24	0.38	0.16	0.12	0.18	-	-	0.47	0.22	0.13
Mean	0.07	0.07	0.09	0.08	0.10	0.07	0.07	0.09	0.07	0.11	0.10	0.09	0.04
STD	0.02	0.02	0.07	0.05	0.09	0.03	0.03	0.03	0.04	0.06	0.11	0.01	

Table C.3: *SL*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	2.28	2.61	-	2.69	2.77	2.98	2.84	2.60	3.33	2.85	2.96	2.79	0.28
2	2.17	2.44	2.97	2.66	2.66	3.00	2.72	2.18	3.23	2.52	2.97	2.69	0.34
3	2.14	2.39	2.82	2.61	2.66	2.94	2.68	2.11	3.19	2.62	3.00	2.65	0.34
4	2.09	2.29	2.75	2.50	2.70	2.87	2.71	2.01	3.14	2.66	2.87	2.60	0.35
5	2.05	2.34	2.73	2.56	-	2.85	2.71	2.01	3.16	2.71	2.80	2.59	0.36
6	2.08	2.30	-	2.50	2.89	2.87	2.67	1.95	3.07	2.49	2.91	2.57	0.38
7	2.09	2.22	-	2.51	-	2.76	2.68	1.94	3.07	2.55	2.84	2.52	0.37
8	2.08	2.21	2.69	2.46	2.85	2.78	2.59	1.87	3.01	2.56	2.89	2.54	0.36
9	2.13	2.04	2.58	2.56	2.86	3.03	2.60	1.89	-	2.62	2.79	2.51	0.37
10	2.04	2.16	2.48	2.51	2.70	2.91	2.63	1.81	3.07	-	2.77	2.51	0.4
11	2.08	2.19	2.46	2.60	2.70	2.92	2.70	1.94	2.99	2.60	2.67	2.53	0.33
12	2.17	2.23	2.63	2.66	2.80	2.82	2.70	1.84	3.02	2.75	2.79	2.58	0.35
13	2.22	2.49	3.03	3.45	3.78	3.31	3.48	2.32	-	-	3.50	3.06	0.58
Mean	2.13	2.30	2.71	2.64	2.85	2.93	2.75	2.03	3.11	2.63	2.90	2.63	0.15
STD	0.07	0.15	0.19	0.26	0.32	0.14	0.23	0.22	0.10	0.11	0.20	0.34	

Table C.4: *STD* of *SL*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	0.03	0.06	-	0.06	0.06	0.1	0.05	0.06	0.14	0.17	0.08	0.08	0.04
2	0.07	0.07	0.06	0.06	0.05	0.1	0.03	0.08	0.07	0.06	0.09	0.07	0.02
3	0.06	0.07	0.09	0.06	0.06	0.08	0.05	0.08	0.07	0.05	0.10	0.07	0.02
4	0.05	0.04	0.08	0.05	0.06	0.06	0.05	0.08	0.04	0.09	0.10	0.06	0.02
5	0.04	0.07	0.08	0.05	-	0.05	0.04	0.08	0.05	0.07	0.09	0.06	0.02
6	0.05	0.09	-	0.04	0.05	0.07	0.05	0.06	0.06	0.05	0.09	0.06	0.02
7	0.05	0.04	-	0.04	-	0.18	0.05	0.05	0.04	0.06	0.10	0.07	0.04
8	0.07	0.06	0.07	0.08	0.05	0.03	0.04	0.09	0.14	0.10	0.11	0.08	0.03
9	0.04	0.07	0.04	0.06	0.04	0.07	0.04	0.07	-	0.07	0.08	0.06	0.02
10	0.07	0.07	0.06	0.05	0.06	0.07	0.05	0.09	0.07	-	0.08	0.07	0.01
11	0.05	0.07	0.06	0.05	0.07	0.06	0.05	0.07	0.06	0.07	0.08	0.06	0.01
12	0.05	0.06	0.06	0.11	0.06	0.05	0.11	0.08	0.07	0.06	0.09	0.07	0.02
13	0.07	0.06	0.13	0.14	0.10	0.15	0.06	0.13	-	-	0.19	0.11	0.04
Mean	0.05	0.06	0.07	0.06	0.06	0.08	0.05	0.08	0.07	0.08	0.10	0.07	0.03
STD	0.01	0.02	0.03	0.03	0.02	0.04	0.02	0.02	0.03	0.03	0.03	0.02	

Table C.5: *SR*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	1.48	1.32	-	1.43	1.37	1.27	1.24	1.32	1.42	1.33	1.46	1.36	0.08
2	1.45	1.26	1.57	1.44	1.39	1.24	1.26	1.30	1.41	1.28	1.49	1.37	0.11
3	1.46	1.25	1.54	1.43	1.40	1.26	1.26	1.30	1.40	1.30	1.48	1.37	0.10
4	1.46	1.23	1.52	1.43	1.40	1.27	1.26	1.30	1.39	1.31	1.48	1.37	0.10
5	1.45	1.22	1.52	1.42	-	1.25	1.26	1.30	1.37	1.32	1.48	1.36	0.10
6	1.43	1.25	-	1.42	1.30	1.25	1.27	1.31	1.36	1.31	1.48	1.34	0.08
7	1.47	1.25	-	1.42	-	1.31	1.27	1.30	1.35	1.31	1.48	1.35	0.08
8	1.46	1.26	1.48	1.39	1.28	1.26	1.27	1.30	1.33	1.32	1.49	1.35	0.09
9	1.44	1.26	1.46	1.41	1.30	1.25	1.26	1.29	-	1.32	1.47	1.35	0.09
10	1.44	1.27	1.43	1.30	1.31	1.24	1.28	1.29	1.34	-	1.47	1.34	0.08
11	1.44	1.28	1.43	1.31	1.30	1.25	1.29	1.32	1.32	1.32	1.45	1.34	0.07
12	1.49	1.27	1.47	1.34	1.28	1.27	1.29	1.29	1.33	1.33	1.48	1.35	0.09
13	1.49	1.32	1.65	1.54	1.40	1.29	1.46	1.31	-	-	1.60	1.45	0.13
Mean	1.46	1.27	1.51	1.41	1.34	1.26	1.29	1.30	1.36	1.31	1.49	1.36	0.09
STD	0.02	0.03	0.07	0.06	0.05	0.02	0.05	0.01	0.04	0.01	0.04	0.04	

Table C.6: *STD* of *SR*

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	0.020	0.024	-	0.016	0.018	0.015	0.013	0.014	0.040	0.026	0.032	0.022	0.009
2	0.026	0.022	0.021	0.021	0.021	0.028	0.014	0.025	0.024	0.021	0.040	0.024	0.006
3	0.025	0.025	0.028	0.026	0.020	0.011	0.009	0.023	0.030	0.015	0.040	0.023	0.009
4	0.022	0.016	0.024	0.018	0.019	0.010	0.012	0.022	0.014	0.015	0.040	0.019	0.008
5	0.018	0.028	0.033	0.025	-	0.018	0.012	0.024	0.018	0.019	0.039	0.023	0.008
6	0.026	0.022	-	0.022	0.023	0.015	0.014	0.025	0.013	0.018	0.042	0.022	0.009
7	0.023	0.015	-	0.022	-	0.087	0.013	0.023	0.016	0.014	0.038	0.028	0.023
8	0.027	0.025	0.038	0.024	0.013	0.011	0.012	0.027	0.017	0.020	0.052	0.024	0.012
9	0.021	0.018	0.015	0.023	0.015	0.013	0.011	0.033	-	0.017	0.043	0.021	0.010
10	0.031	0.023	0.027	0.018	0.015	0.019	0.011	0.034	0.022	-	0.034	0.023	0.008
11	0.023	0.018	0.033	0.014	0.017	0.011	0.016	0.035	0.021	0.017	0.028	0.021	0.008
12	0.023	0.014	0.031	0.025	0.016	0.021	0.011	0.035	0.015	0.017	0.036	0.022	0.009
13	0.027	0.013	0.064	0.040	0.080	0.021	0.019	0.038	-	-	0.083	0.043	0.027
Mean	0.024	0.020	0.031	0.023	0.023	0.022	0.013	0.027	0.021	0.018	0.042	0.024	0.001
STD	0.003	0.005	0.013	0.006	0.019	0.020	0.003	0.007	0.008	0.003	0.013	0.009	

Table C.7: Strides

Lap	R1	R2	R3	R4	R5	R6	R7	R8	R9	R10	R11	Mean	STD
1	37	32	-	32	30	28	29	24	29	-	28	30	4
2	39	35	30	35	31	18	30	30	30	-	28	31	6
3	39	35	31	35	31	17	31	32	31	29	27	31	6
4	40	37	33	37	30	16	31	34	31	33	29	32	6
5	40	36	33	36	-	14	31	34	31	32	30	32	7
6	40	36	-	36	29	14	31	35	31	31	29	31	7
7	40	37	-	37	-	14	31	35	31	30	29	32	8
8	40	38	17	38	29	13	32	37	32	33	28	31	9
9	39	40	17	40	29	12	32	35	32	33	30	31	9
10	40	38	16	38	31	11	32	38	32	32	30	31	9
11	40	38	15	38	21	11	31	35	31	32	31	29	10
12	38	37	16	37	19	11	30	37	30	-	30	29	10
13	37	33	17	33	22	11	23	28	23	32	23	26	8
Mean	39	36	23	36	28	15	30	33	30	32	29	30	2
STD	1	2	8	2	5	5	2	4	2	1	2	2	