The effect of fatigue on leg- and joint stiffness during running

Bachelor Thesis Biomedical Engineering

Freek Bolding
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Supervisors:
dr. J. Reenalda
R.P. van Middelaar
dr.ir. A.Q.L. Keemink

Faculty of Electrical Engineering,
Mathematics and Computer Science
Biomedical Signals and Systems
University of Twente

Abstract

Background: While running is associated with various health benefits, incidence of injuries is high. The vast majority of injuries are related to overuse. Shock attenuation and the related stiffness of the leg and hip, knee and ankle joints have been implied as factor in overuse injuries. Fatigue is thought to affect the lower extremity stiffness, possibly increasing the risk of overuse injuries in this manner. Therefore, this study aimed at determining the effect of fatigue on the leg, hip, knee and ankle stiffness.
Methods: For this purpose, a data set consisting of nine subjects running in both a non-fatigued and a fatigued state was analyzed. Motion was captured with a VICON optical system tracking 34 markers on the body of the subject. Ground reaction forces were measured with two force plates. OpenSim, a software system designed for biomechanical simulation and analysis, was used to perform inverse kinematic and inverse dynamic analysis. From the resulting joint angles and moments, the leg, hip, knee and ankle stiffnesses were calculated.
Results: Subjects, when running fatigued, showed a significant reduction of the leg stiffness and a significant increase of the knee stiffness when compared to non-fatigued running. The hip stiffness could not be suitably determined, as no linear relation between the net hip moment and hip angle was found between initial contact and mid-stance. The ankle stiffness did not show significant changes between non-fatigued and fatigued running.
Conclusions: A decrease of the leg stiffness and a increase of the knee stiffness can both be associated with overuse injuries, implying running while fatigued increases the odds of sustaining an overuse injury. Deeper analysis of the changes in ground reaction forces, leg compression, joint moments and joint angles is needed to better understand the found stiffness changes. Future studies are recommended to use a treadmill with integrated force plates to increase the quantity and consistency of analysed strides.

***Keywords*—** Running, Fatigue, Leg stiffness, Joint stiffness, Quasi-stiffness, OpenSim

Contents

1 Introduction

Running has been gaining popularity since the running boom in the 1970s, as running in public became more accepted [1]. Physical exercise in public was mostly frowned upon before then, which meant that running was limited to organised races. With changes in public opinion and manners, recreational jogging on the streets became increasingly common [2]. This change was in part due to the public becoming aware of the substantial health benefits of running. These benefits include, among others, an increase in physical fitness and a reduced risk of obesity, cardiovascular disease and other chronic health problems [3, 4].

However, running is also associated with injuries. Incidence of injuries in the lower extremities is high, with numbers ranging from 19.4% to 79.3% [3]. This large variation is due to differences in study populations and definitions of injury [5]. The vast majority of injuries are related to overuse, especially for novice runners [6]. Overuse injuries are caused by repeated applied stresses with inadequate rest time between the stress applications [7]. Risk factors for running-related injury are numerous and vary a lot. Examples of risk factors are run distance, frequency, shoe design, terrain, fatigue, age, experience and a history of previous injury [8]. These factors are also dependent on the type of injury and which body part it affects. Different types and locations of injuries are associated with varying risk factors. The most common location for injury is the knee, with around 41% of the cases [3, 6]. Similarly, the most common running injury, which is patellofemoral pain syndrome, also affects the knee [9]. One reason for the large amount of knee injuries may be the fact the knee absorbs a large portion of the impact forces and is the main shock absorber of the body [10].

These impact forces, which can be more than twice the body weight during running [11], cause a shock wave of energy throughout the body. These shocks have been implicated in overuse injuries [12, 13]. Dissipation of the shocks while they travel up the body is called shock attenuation [14]. Although the precise relation between shock attenuation and injuries is unclear [15], shock attenuation is, for example, suggested to be a factor in the development of tibial stress fractures [16]. It is suggested that when fatigue develops during running, shock attenuation is decreased and becomes more dependent on passive mechanisms [17, 18].

Shock attenuation is related to the lower body's mechanical stiffness [19]. Stiffness is defined as the extent to which the legs or lower extremity joints resist deformation upon contact with the ground [20]. Therefore, a stiffer leg compresses less when subjected to a particular force while a stiffer joint rotates less when subjected to a particular moment. Stiffness is required for running performance, but is also associated with injuries [21]. While multiple studies have been done on the effect of fatigue on leg stiffness during running [22, 23, 24, 25], only two studies were found on the effect of fatigue on joint stiffness for rearfoot strike runners. Both these studies, by Weir et al. and Luo et al., only tested male subjects and were done on a treadmill [26, 27]. Therefore, both studies warned for reduced generalizability of their findings for female runners. Additionally, Weir et al. did not take into account the hip stiffness [26]. The aim of this thesis is to find the effect of fatigue on lower extremity leg and joint stiffness.

For the remainder of this section, the parameters shock attenuation, fatigue and stiffness will be further clarified. Additionally, OpenSim, which is an open-source software system to simulate dynamic movement, will be described and at last the research question and hypothesis will be stated.

1.1 Shock attenuation

The ground reaction forces cause a shock wave of energy throughout the body. The magnitude of these forces during contact with the ground can be seen in figure 1. The loading rate, which is the speed at which forces impact the body, is also visible in figure 1. The shock waves travel mostly through the skeletal system and are dissipated along the way [14]. It is suggested that these shocks, in combination with the loading rate, are a factor in the development of injuries [16]. The absorption of impact energy and reduction of the amplitude of the shock wave is called shock attenuation [28]. This leads to a much reduced vertical acceleration of the head on foot impact with the ground compared to the tibia [15]. Shock can be attenuated by both passive and active mechanisms [29]. Passive shock attenuation is done by anatomical structures such as the heel pad, ligaments, articular cartilage and bone [12]. Active shock attenuation is done by eccentric muscle contraction and changes in joint configuration [19]. It is suggested that when fatigue develops in the runner, shock attenuation decreases and becomes more dependent on passive mechanisms [17, 18].

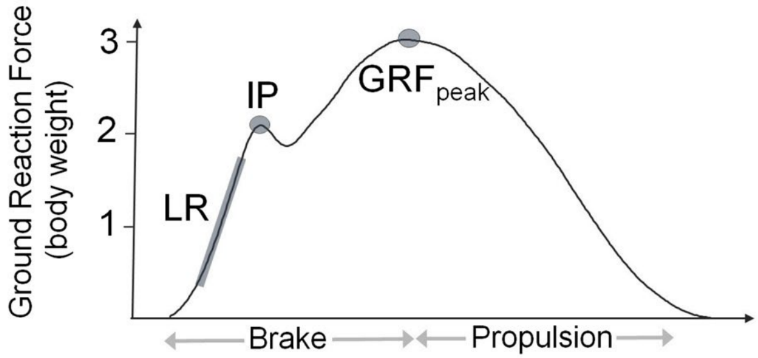

Figure 1: Ground reaction force during stance phase [30]. The peak ground reaction force (GRF_{peak}) is during mid-stance. LR shows the loading rate and IP shows the impact peak.

1.2 Fatigue

Fatigue is a non-specific symptom associated with many conditions. There are various forms of fatigue, which can be classified as either mental or physical fatigue. Mental fatigue refers to the cognitive or perceptual aspects of fatigue, such as difficulty in planning, concentrating [31]. Physical fatigue refers to the (decreased) performance of the motor system [32]. The latter is the interest of this study. Different kinds of physical fatigue exist, like cardiovascular fatigue which is related to the oxygen delivery to the muscles [33]. Muscle fatigue, which is another form of physical fatigue, is defined as a decrease in force generation by the muscles [18]. Its mechanism is complex as it is influenced by the concentrations of many different factors: neurotransmitters, calcium ions, oxygen, ATP, metabolic factors and fatigue reactants [32]. There is not one single cause of fatigue, but rather a combination of mechanisms that causes it. One example of such a mechanism is the decrease of motor neuron firing rates after repeated activation due to decreased excitability to synaptic input [34].

Whether fatigue is induced by a lack of ATP, oxygen or any other possible cause, the result is a reduction in maximum muscle force. This makes it harder for the muscles to actively absorb the impact shocks when landing on the ground by eccentric muscle contraction. Changes in running characteristics as a response to fatigue are widely different between individuals [23]. For example, some runners decrease their stride frequency while others were found to increase it [23, 25]. These variations in individual responses, in combination with differing running protocols, lead to studies finding contradicting results for the effects of fatigue on running characteristics [22]. Another example is the peak vertical ground reaction force (vGRF). This is the maximum reaction force exerted by the ground when the leg lands during the stride cycle. Some studies [13, 22] found an increase in peak vGRF while others [24, 35] found it had decreased. These variations make it difficult to make general and concise statements about the effect of fatigue on running.

The leg stiffness, which is a central part of this study and will be elaborated further on in section 1.3, was generally found to decrease [22, 25, 35, 36] although some studies found no significant change [23, 24]. In ultra-long distance events the leg stiffness was found to increase [37, 38, 38], but these events are not within the scope of this thesis. Recently, Zandbergen et al. did a literature review on kinematic changes caused by fatigue. This review found a decreased lower body stiffness, an increase in knee flexion, increased shock attenuation and increased peak accelerations at the tibia and sacrum but not the head [39]. Interestingly, an increase in vertical center of mass displacement was found for recreational runners, but it was decreased for experienced runners. Joint moments and joint stiffness were not taken into account for this review, as not enough studies with relevant findings were found. A recent study by Weir et al. found an increase in knee stiffness and decrease in ankle stiffness during prolonged running, although a fatigued state was not necessarily reached and leg stiffness was not changed [26]. The effect of fatigue on hip stiffness was not found in existing literature. Yu et al. found no change in the hip angles, but did find a reduction of the peak hip moment after fatigue [40]. The joint angles and moments are important parameters ultimately determining the joint stiffness.

1.3 Stiffness

One aspect that determines the shock attenuation is the stiffness [19]. The stiffness describes the relationship between the deformation of a body and a given force [21]. The true stiffness of the leg is the combination of the individual stiffness values of the tissues in the body (e.g. muscles, tendons, bone, ligaments, cartilage) [41]. Taking all these factors into account separately is extremely complex and not feasible [26]. Therefore it is simplified by using spring-mass models

to represent the leg and joints. It was found that reduced leg stiffness was associated with increased shock attenuation [42]. However, stiffness is also necessary for performance [21]. And while an increased leg stiffness may induce bony type injuries, a decreased leg stiffness can lead to soft tissue-type injuries by allowing for excessive joint motion [26, 43]. Furthermore, Messier et al. found an increased knee stiffness to be the sole predictor of running related injury [44].
To make meaningful statements about leg stiffness and joint stiffness, these parameters need to be precisely defined. Stiffness as a concept has its origin in Hooke's law [21]. For linear springs, Hooke's law is defined as

$$F = kx \tag{1}$$

where F is the force (in Newton) required to move a mass on a spring with spring constant k (in Newton meter^{-1}) with distance x (in meters). Hooke's law also has an angular form, which is used for torsion springs. This angular form is defined as

$$\tau = \kappa\theta \tag{2}$$

where τ is the moment (in Newton meter) required to twist a torsion spring with torsional spring constant κ (in Newton meter degree^{-1}) by angular distance θ (in degrees). Equation (1) and (2) will be used for calculation of leg stiffness and joint stiffness, respectively.

1.3.1 Leg stiffness

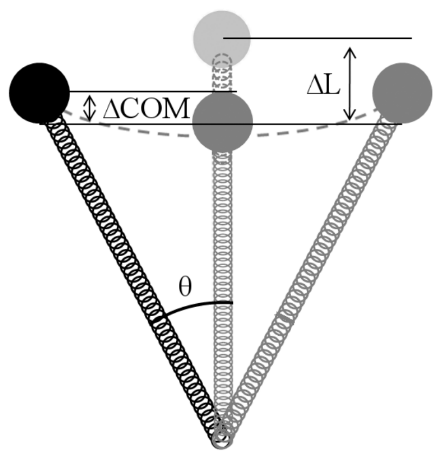

Figure 2: Spring-mass model of the leg during running [20]. The body is represented as a point mass located at the center of mass (COM) which is connected to the foot by a single spring, representing the leg. The model shows three phases of a stride; when the foot touches the ground (left), during midstance (middle) and as the foot leaves the ground (right).

The model to calculate 'leg' stiffness for running was first developed by McMahon and Cheng [45]. In their model, the leg was regarded as having the properties of a simple spring. When the leg is regarded as a massless spring loaded by the mass of the body, the spring constant of Hooke's law can be regarded as the stiffness of the leg [46, 47]. Figure 2 shows this spring-mass model. The leg spring can be seen hitting the ground at a certain angle θ. If the leg would be rigid, the leg would not be compressed and the center of mass would be located at the light grey circle at the top of the figure. However, the center of mass is found to be lower at mid-stance than at initial contact, meaning the leg has compressed. The amount of compression is denoted as ΔL. The ratio between the maximum vertical ground reaction force and the leg compression is defined to be the leg stiffness [46].

1.3.2 Joint stiffness

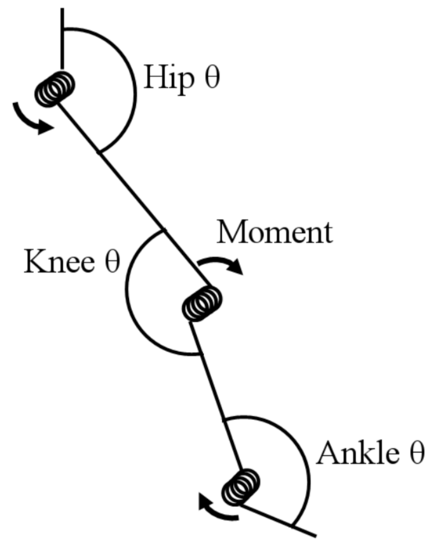

Figure 3: Torsion spring model of the lower limb joints [20]. θ shows the angle of the joints while the arrows show the moment acting on them. The direction of the arrows shows the direction of a positive defined joint moment. The angles and moments of interest are those belonging to the sagittal plane.

This leg compression is obtained by rotating the hip, knee and ankle joints, which dissipates the impact forces and stores energy for elastic return in the next step [20]. The stiffness of these joints therefore offers an insight into shock attenuation strategies and relative loading at each joint. The torsional stiffness of a joint is usually defined as the ratio between the net joint moment and the angular displacement of that joint [48]. However, this definition has been deemed 'quasi-stiffness' by Latash et al., as it is a distinct concept from the actual stiffness in the context of powered joints [41, 49]. As stated by Rouse et al., quasi-stiffness is a description of the dynamic task in the moment-angle domain, rather than a true representation of stiffness or impedance [49]. Quasi-stiffness is the ability of an system to resist externally imposed displacements, which is also determined by the system's inertial, viscous and elastic elements [41, 50]. A model that could be able to determine the true joint stiffness, taking into account all different elements and additionally the multiple muscles, some of which are bi-articular, would become too complex [21]. Therefore, the simpler quasi-stiffness will be used for this study, which is simply the slope of of the linear fit to the moment-angle curve of a joint in a specific task [51]. The quasi-stiffness will be called stiffness for the remainder of this thesis.

Figure 3 shows the torsion spring model of the lower extremity. The flexion angle of the hip and knee and dorsiflexion angle of the ankle are defined as the positive angles. The extension moment of the hip and knee and plantar flexion moment of the ankle are defined as the positive moments. The joint stiffness determines how much the joint angle changes in response to a given external moment. When the stiffness of the joints is greater, the joint angle changes less during contact which results in less leg compression and higher leg stiffness [48].

1.4 OpenSim

OpenSim is an open-source software system designed to let users develop musculoskeletal models and perform dynamic simulations of movements. It allows users to study neuromuscular coordination, analyze athletic performance and even identify sources of pathological movement [52]. It is fitted with various tools like musculoskeletal model scaling, forward and inverse dynamic analysis and computed muscle control to compute muscle excitations. OpenSim is useful for the determination of the joint stiffnesses, as it can provide the kinematics and kinetics of running based on motion capture and ground reaction force data. The kinematics, which include joint angles, and kinetics, which include net joint moments, are obtained from inverse kinematic and inverse dynamic calculations respectively. The inverse kinematic problem is solved by minimizing the root-mean-square error between the measured marker locations and the virtual markers of the musculoskeletal model, while taking joint constraints into account [52]. The inverse dynamic problem is solved by defining and solving motion equations, essentially calculating the net joint moments required to move the joints as determined by the inverse kinematic analysis, taking mass distribution and ground reaction forces into account.

1.5 Research question

Taking into account the aims of the study, the following research question was developed: *How do the leg stiffness and the hip, knee and ankle (quasi-)stiffness change as a result of running-induced fatigue?* Given the results of prior studies [22, 35, 39], it is hypothesized that the leg stiffness will be reduced in fatigued state. The ratios between the joint stiffnesses are expected to change as fatigue develops, specifically by an increase of the knee stiffness and a decrease of the ankle stiffness. This expectation is based on findings of Weir et al. and Luo et al. [26, 27]. The hip stiffness is expected to decrease, as peak hip moment was found to decrease with fatigue [40].

2 Methods

The used data set was collected for an earlier thesis and was therefore already available for this study. No additional experiments were done, due to regulations induced by the current Covid-19 pandemic.

2.1 Subject information

The data set consists of nine subjects (*5 males, 4 females, 28.2 ± 10.1 yrs, 180.6 ± 9.7 cm, 71.4 ± 9.6 kg*). The subject characteristics can be found in table 1. All subjects ran recreationally and ran a minimum average of 10 kilometres per week for at least a year. All subjects were rearfoot strikers. This was validated by recording the lower body with a handheld camcorder (Sony DCR-SX45E, Sony, Tokyo, Japan) while the subjects ran at a preferred speed. Their foot strike pattern was analysed and showed the rear end of the shoe touching the ground first. The subjects reported no injuries in the past six months. The dominant leg was defined as the leg the subject would use to kick a ball. The local ethics committee (METC Twente) approved the experimental protocol of this study. All participants signed informed consent.

Table 1: Subject characteristics

Subject	Sex	Mass (kg)	Height (cm)	Age (years)	Dominant leg
1	Male	60.0	174.0	24	R
2	Male	74.8	184.0	25	R
3	Male	84.9	194.5	21	R
4	Male	80.6	192.0	21	R
5	Female	74.9	172.5	23	L
6	Female	63.6	183.5	26	R
7	Male	79.3	187.0	25	R
8	Female	69.3	175.0	55	R
9	Female	54.9	162.5	34	R
Mean	-	71.4	180.6	28.2	-
SD	-	9.6	9.7	10.1	-

2.2 Experiment protocol

The subjects warmed up with a self-chosen method. After the warm-up, the protocol started. It consisted of three parts: non-fatigued overground running, treadmill running to fatigue and fatigued overground running. In the first part, subjects ran back and forth on a 10-meter runway. The subjects were asked to run 10 or 12 km/h, dependent on the velocity the subject was going to run. The run velocity was regulated by a metronome which gave a sound if the subject needed to be on either side of the runway. The subject had two seconds to turn around and start the next run. Each run from one side to the other was called a 'trial'. The objective of each trial was to land on one of two force plates with either the left or right foot. A trial was deemed successful if the whole foot landed within the boundary of a force plate. This was continued until five successful trials were achieved for both feet or when the subject had tried 40 times.

The goal of the second part of the protocol was to rapidly induce fatigue. The subjects ran on a treadmill with a speed of 103% of their average speed for an 8-km race, until they felt they could only continue for two more minutes to finish the complete protocol. This was supported by a heart rate increase of at least 15% compared to the heart rate after 1 minute of the fatiguing run. Additionally, every 3 minutes a revised Borg scale was shown, which is a subjective measure of tiredness on a 1-10 scale with 8 or higher indicating fatigue. The first part of the protocol was then repeated for the fatigued overground running.

2.3 Measurement systems

Motion was captured with a VICON optical system (VICON Nexus 2.10, VICON Vantage, Oxford, UK, eight cameras). High-speed infrared cameras track 34 reflective markers on the lower body of the subject. The markers were placed on anatomical landmarks and as cluster markers on the foot, tibia and thigh. The locations of the markers on the body can be seen in figure 4. The VICON system then correlates the data from each camera to generate a three-dimensional map of all markers. The system was calibrated as prescribed by VICON. Static calibration was done by having the subject stand still in anatomical position on the force plate. VICON software (VICON Nexus 2.10) was used to acquire data.

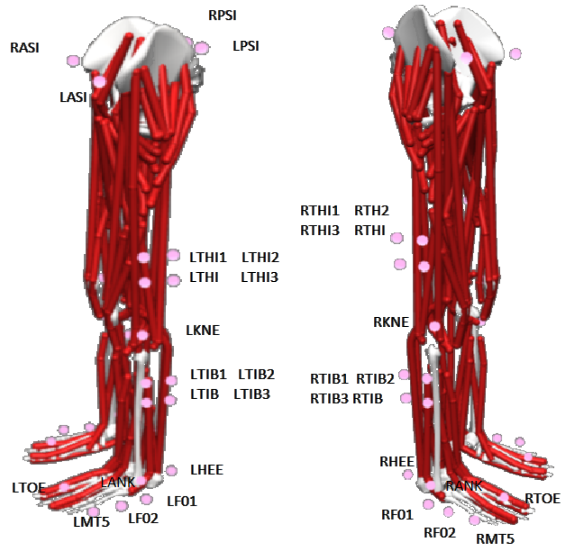

Figure 4: Locations of reflective VICON markers on a musculoskeletal model, with corresponding marker names. The model represents the muscles and body segments of the subject's lower body, and is used to perform inverse kinematics and inverse dynamics in OpenSim. Courtesy of Robbert van Middelaar

Three-dimensional ground reaction forces were recorded by two AMTI OR6 Series force plates (AMTI Force and Motion. Watertown, MA, USA, 1000 Hz). VICON data (100 Hz) were interpolated in MATLAB to match the 1000 Hz force plate data. The fatigue-inducing run was performed on a treadmill (C-Mill, Motek Medical, Culemborg, the Netherlands) and heart rate was measured during the fatigue-inducing run with a heart rate band around the chest (Polar RS400sd, Polar Electro, Kempele, Finland).

2.4 Data pre-processing

The collected data from the VICON system and the force plates, henceforth referred to as motion and GRF data respectively, were saved into .mat files. Each file contained the data from either fatigued or non-fatigued running of a specific subject. The collected motion and GRF data were first split by trial. As the field-of-view of the VICON was limited and did not extend to the edge of the runway, marker coordinates were given as zeros when the subjects were on either end of the runway. This was used to split each trial. The trials were determined by finding each frame where every marker had a nonzero vertical coordinate. The vertical coordinates were used as they were always nonzero when the marker could be observed by the optical system. Using this, the start- and endpoints of each trial were identified. All motion and GRF data belonging to the relevant trials were selected with these start- and endpoints.

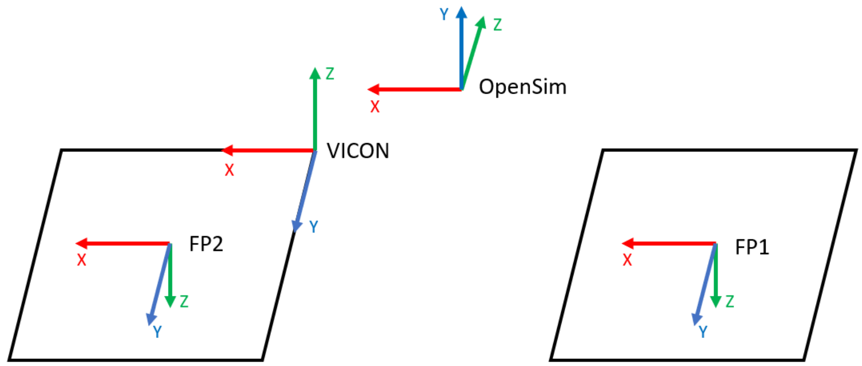

Figure 5: The different coordinate systems of the VICON system, force plates (FP) and the OpenSim software. In addition to having different origins, the direction of the vertical axis of the FP system is reversed compared to the vertical axis of the VICON system. The OpenSim is rotated 90° around the x-axis compared to the VICON system.

The motion and GRF data are defined in different coordinate system. Additionally, the coordinate system used in the OpenSim software is oriented in another way as well. This is shown in figure 5. Therefore, the motion data was first translated to the origin of either force plate 1 or force plate 2, depending on which plate was stepped on in that particular trial. The GRF z-axis data was flipped to match the z-direction of the motion data. As the subjects ran back and forth, half the trials involved a negative horizontal velocity. These trials were rotated 180° around the VICON z-axis to get every trial in the same direction. The motion and GRF data were then rotated +90° around the x-axis to get the orientation required for the OpenSim software. The ground reaction forces and moments were filtered with a fourth-order low-pass Butterworth filter with a cut-off frequency of 20Hz. The stance phase, between initial contact and toe-off, was determined based on where the vertical ground reaction force exceeded a 33 N threshold. This threshold was chosen to prevent measured vibrations from causing incorrectly determined contact times. Measured forces and moments outside these contact times were set to zero.

2.5 Leg stiffness

Using equation 4, the leg compression was then calculated. Equation 3 was then used to calculate the leg stiffness for each trial.

With the spring-mass model from figure 2, the following equation is used to calculate leg stiffness (in Newton meter^{-1})

$$k_{leg} = \frac{F_{max}}{\Delta L} \tag{3}$$

where F_{max} is the maximum value of the vertical ground reaction force (in Newton) and ΔL is the change in vertical leg length (in meters) [46]. The maximum vertical ground reaction force was easily obtained from the GRF data, but the change in leg length (compression) needs to be estimated. This can be done in multiple ways [46]. In 2012, Coleman et al. compared multiple different ways of estimating the compression of the leg to true measurements and found the following method first proposed by Morin et al.[53] to be the most similar to direct measurement results [54]. With this method, which was used in this study, ΔL was derived from

$$\Delta L = L_0 - \sqrt{L_0^2 - (\frac{v t_c}{2})^2} + \Delta COM \tag{4}$$

where L_0 represents the initial leg length (in meters), which is defined as the distance from the greater trochanter to the ground while standing upright, v represents running velocity (in meter second^{-1}), t_c represents ground contact time (in seconds) and ΔCOM represents the vertical height change of the center of mass during ground contact (in meters). Running speed was calculated by averaging the x-coordinates of the four hip markers at the start and end of a trial. The difference between these two averages was then divided by the elapsed time to get an averaged running speed. The vertical height change of the center of mass was estimated by averaging the y-coordinates of the four hip markers during the stance phase and subtracting the lowest value from the highest. It was assumed the change in height of the center of mass was similar to the change in height of the hip.

2.6 Joint stiffness

As stated in section 1.3, the (quasi-)stiffness of the joints is defined as the ratio between the net muscular moment and the angular displacement of the joint. These joint angle and moment are obtained from inverse kinematic and inverse dynamic analysis respectively. For this purpose, OpenSim is used to simulate the dynamic movement of a musculoskeletal model. To generate simulations, OpenSim requires two types of data files: Marker trajectories (.trc) and external load data (.mot) consisting of ground reaction forces, centers of pressure and free moment data. The centers of pressure were calculated with the following equations [55]

$$CoP_x = (M_z - F_x \cdot y)/F_y \tag{5}$$

$$CoP_z = (-M_x - F_z \cdot y)/F_y \tag{6}$$

where CoP_x and CoP_z represent the x- and z-coordinates of the centers of pressure (in meter), M_x and M_z represent the x- and z-components of the measured ground reaction moment (in Newton meter), F_x, F_y and F_z represent the measured ground reaction force components (in Newton) and y represents the distance from the surface of the force plate to the force transducers (in meter). This distance was already compensated for by the force plate and is therefore equal to zero. Additionally, the so called free moment was calculated as well. This free moment gives the moment about the y-axis when the total reaction forces and moments are replaced by a single force vector originating at the center of pressure. The free moment is calculated with the following equation [55]

$$FM = M_y - F_x \cdot CoP_z + F_z \cdot CoP_x \tag{7}$$

where, in addition to the already established variables, FM represents the free moment (in Newton meter) and M_y represents the y-component of the ground reaction moment (in Newton meter) which was measured by the force plate. The motion data was then written to .trc files and the ground reaction forces combined with the centers of pressure and free moment were written to .mot files for each trial.

These files then served as input for OpenSim. Data processing in OpenSim consisted of three phases. First, the model was scaled to the body of the subject, using OpenSim's scale tool. This scaling used the static pose markers and the subject's mass as input. Model scaling was an iterative process, where markers had to be moved on the model to better match the marker locations on the subject. Scaling was done iteratively until the root-mean-square error was smaller than 1 cm. The marker trajectories were then used to perform inverse kinematic analysis on the scaled model. The inverse kinematic analysis returns the joint angles over time, or generalized coordinate trajectories. The generalized coordinate trajectories in combination with the external load data were used to perform an inverse dynamic analysis on the scaled model. The inverse dynamic analysis calculated net joint moments by defining and solving motion equations. Figure 6 shows the steps taken in OpenSim to obtain the joint angles and moments.

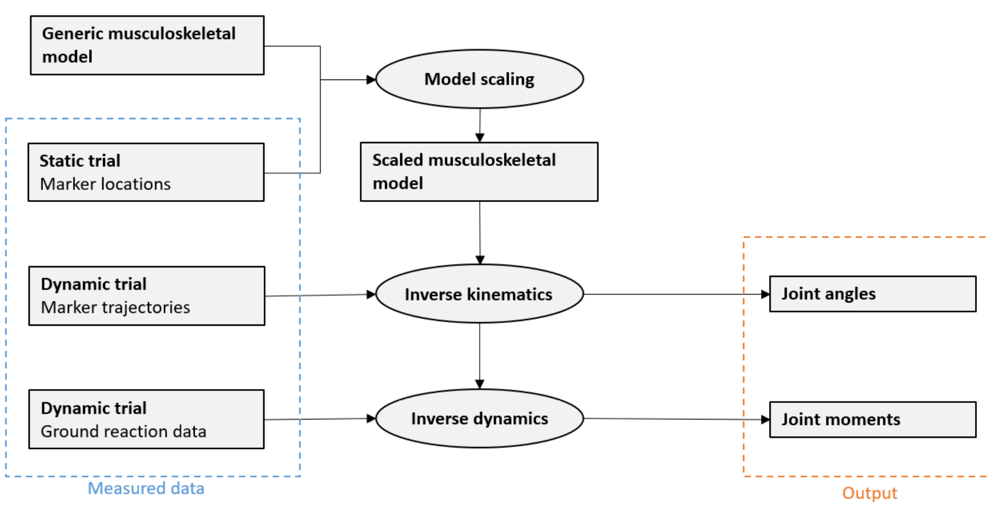

Figure 6: Workflow in OpenSim software.

The joint angles and moments were opened in MATLAB using the function `importdata`. Per trial, the hip, knee and ankle angles and moments of the leg stepping on the force plate were saved. Mid-stance was defined as the point in the stance phase where knee flexion was maximal. The joint stiffness is calculated using following equation

$$k_{joint} = \frac{\Delta M}{\Delta \theta} = \frac{M_{MS} - M_{IC}}{\theta_{MS} - \theta_{IC}} \tag{8}$$

where M_{MS} , θ_{MS} , M_{IC} and θ_{IC} are the joint moments and angles at mid-stance and initial contact respectively. This is shown in figure 7. As the joint stiffness is determined using only these two points, it is the average stiffness between initial contact and mid-stance. Because for some subjects the ankle dorsiflexion angle first decreased at the initial contact before increasing, the stiffness of the ankle was calculated slightly different. Instead of using the moment and angle at initial contact, the minimum ankle angle and its corresponding moment were used for the joint stiffness calculation. For the subjects where the ankle angle was already minimal at initial contact this alteration made no difference, while for the other subjects the calculated stiffness better approached the linear fit between the moment and angle.

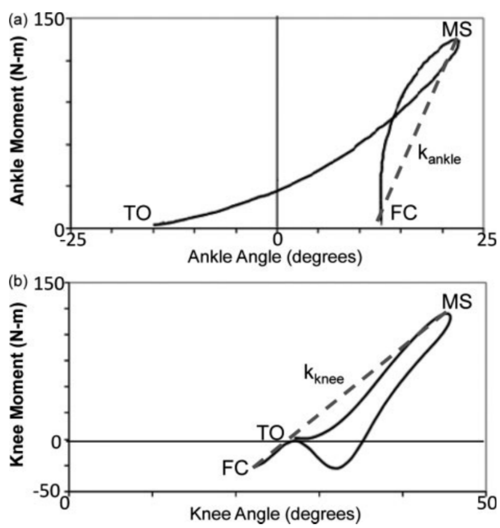

Figure 7: Visualization of how the joint stiffness is determined from the joint moment-angle graph between initial contact (FC) and mid-stance (MS) [56].

2.7 Data analysis

After the leg- and joint stiffnesses were calculated for every trial, the data was analyzed. Only the trials concerning the subject's dominant leg were investigated. The means and standard deviations of the four different stiffnesses (leg, hip, knee, ankle) were calculated for the before and after fatigue phase for all subjects. Additionally, two other values were determined: the sum of the knee and ankle stiffnesses, $k_{Knee} + k_{Ankle}$, and the ratio between them, k_{Knee}/k_{Ankle}. The knee and ankle combination was found to provide the best correlation with running and to increase the reliability of the results when compared to joint stiffnesses on their own [20]. For $k_{Knee} + k_{Ankle}$, the propagated uncertainty was calculated by simply summing the standard deviation of both stiffnesses. For k_{Knee}/k_{Ankle}, the propagated uncertainty was calculated by first summing the relative standard deviations of both stiffnesses and then multiplying it with the corresponding k_{Knee}/k_{Ankle} value.

To compare before and after fatigue, a paired, two-tailed Wilcoxon signed rank test was used in MATLAB [57]. This non-parametric test gives the P-value of the null hypothesis that the difference between the stiffness before and after fatigue has a zero median. The null hypothesis (H0) is rejected when the P-value is smaller than 0.05, implying a 5% significance level. Variability of the results can be illustrated with the coefficient of variation. The coefficient of variation was calculated with

$$CV = \frac{\sigma}{\mu} \cdot 100\% \tag{9}$$

where σ is the standard deviation (or propagated uncertainty) and μ the mean stiffness value. A subject was defined as showing high variability for a specific stiffness calculation when the coefficient of variation was greater than 20%. This boundary was arbitrarily chosen. Section 4.2.2 in the Discussion will delve deeper into the cause of a large coefficient of variation.

3 Results

In this section the results per subject will be shown to display information about the leg stiffness and the stiffnesses of the joints during the situations before and after fatigue. As mentioned in section 2.7, only the trials concerning the dominant leg were investigated. A table containing the exact stiffness values, their standard deviation and coefficient of variation can be found in appendix A.1, see table 2.

3.1 Leg stiffness

The leg stiffness and corresponding standard deviation of each subject before and after fatigue is shown in figure 8. The leg stiffness changed significantly when comparing before and after fatigue ($P = 0.0117$). Every subject, with exception of subject 5, showed a reduced leg stiffness after fatigue was induced. No subject showed high variability.

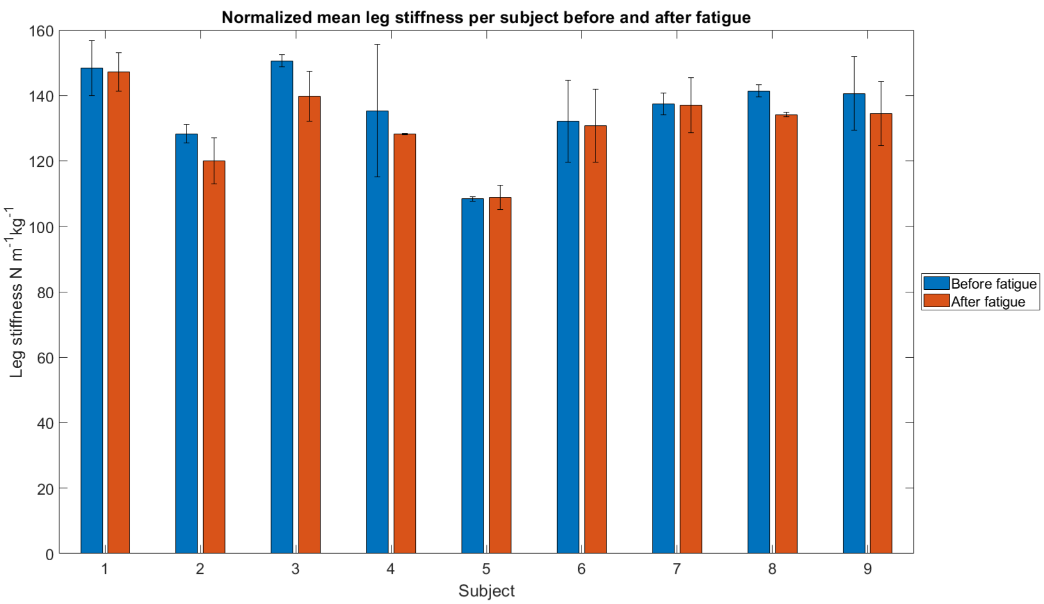

Figure 8: The mean leg stiffness and its standard deviation for each subject before and after fatigue. The values are normalized by dividing by the subject's mass.

3.2 Joint stiffness

Figure 9 shows, as an example, the joint angles and moments between initial contact and mid-stance for one subject and phase. The data from the three analyzed trials of this particular subject and protocol phase is plotted, with each trial being a separate line. These graphs for the other subjects can be seen in appendix A.2. The three graphs on the bottom row show the relation between the joint moment and joint angle. The knee and ankle show a linear relation between the joint moment and angle, of which the slope is the joint's stiffness. The hip lacks a linear relation between the joint moment and angle.

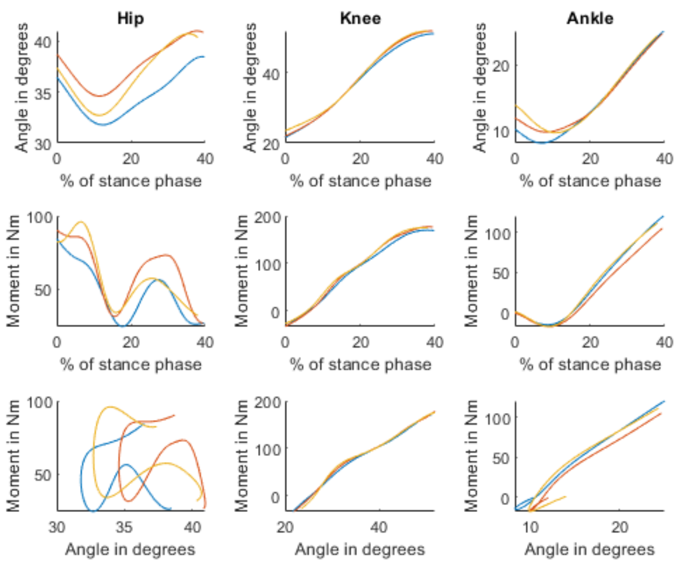

Figure 9: The joint angles and moments between initial contact and mid-stance of subject 5, before fatigue. The left, middle and right column show the data from the hip, knee and ankle respectively. Top row shows the angle of each joint, while the middle row shows the moment. The bottom row shows the moment plotted against the angle. Flexion angle of the hip and knee and dorsiflexion angle of the ankle were defined as the positive angles. Extension moment of the hip and knee and plantar flexion moment of the ankle were defined as the positive moments.

The hip stiffness of each subject before and after fatigue is shown in figure 10. The hip stiffness shows highly variable results, with both positive and negative values. All subjects showed high variability and for some subjects the standard deviation was more than two times the actual stiffness value. In addition, some subjects showed very large hip stiffnesses. For example, the hip stiffness of subject 3 after fatigue was $-1.5 \cdot 10^4 Nm/deg$. The y-axis of the hip stiffness bar graph was therefore limited to $\pm 1 Nmdeg^{-1}kg^{-1}$ to keep the results of the other subjects visible. The Wilcoxon signed rank test indicated a significant change of the hip stiffness after fatigue was induced ($P = 0.0391$). The hip stiffness will be extensively discussed in the Discussion (section 4.2.1).

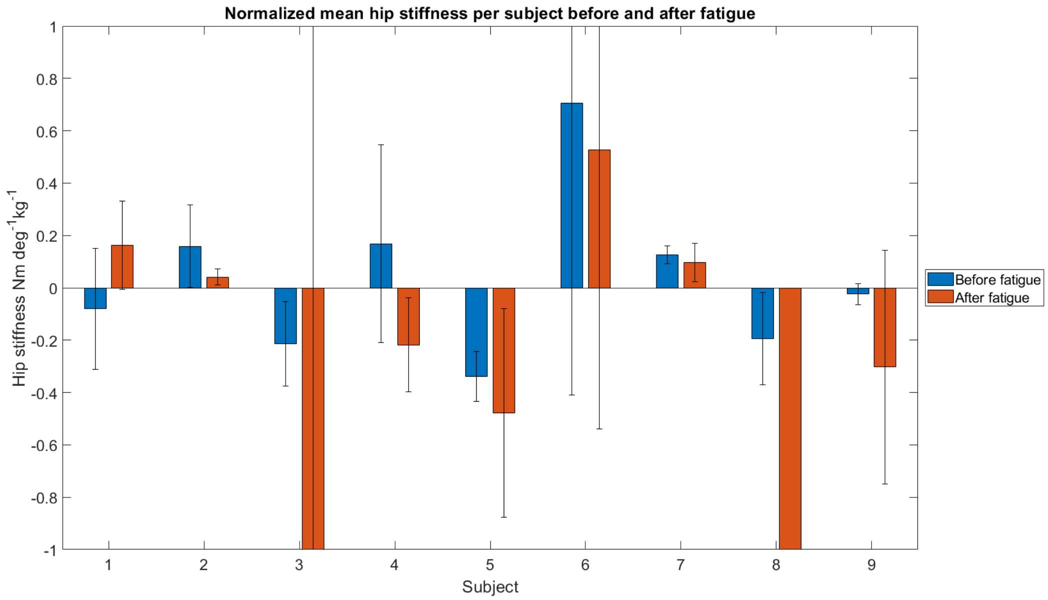

Figure 10: The mean hip stiffness and its standard deviation for each subject before and after fatigue. The values are normalized by dividing by the subject's mass.

The knee stiffness of each subject before and after fatigue is shown in figure 11. The knee stiffness changed significantly when comparing before and after fatigue ($P = 0.0195$). Subjects 4 and 7 show high variability before fatigue, with subject 7 showing high variability after fatigue.

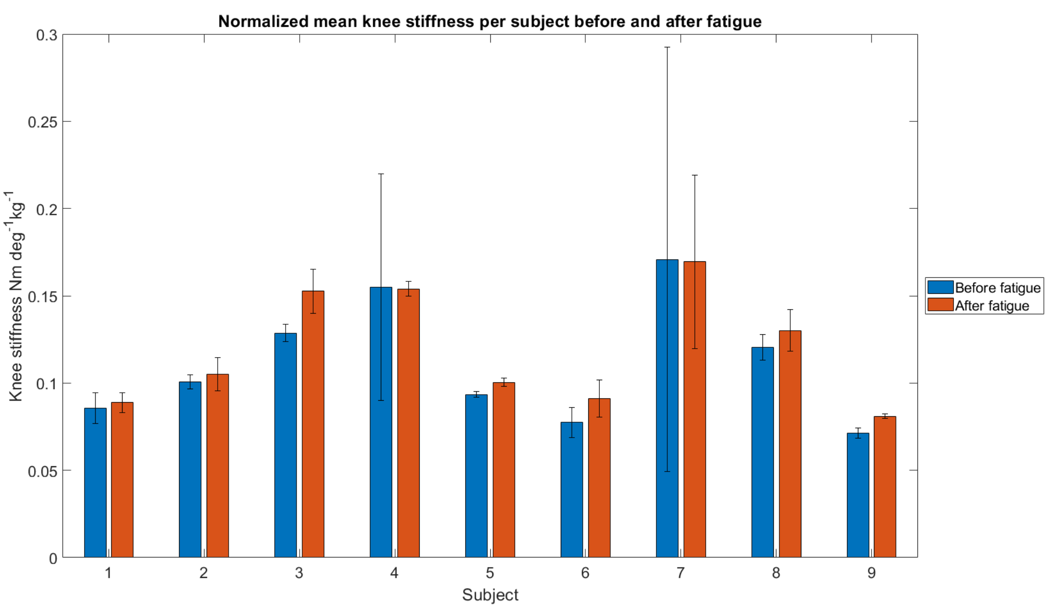

Figure 11: The mean knee stiffness and its standard deviation for each subject before and after fatigue. The values are normalized by dividing by the subject's mass.

The ankle stiffness of each subject before and after fatigue is shown in figure 12. The ankle stiffness did not change significantly when comparing before and after fatigue ($P = 0.7344$). Subjects 4 and 7 show high variability before fatigue.

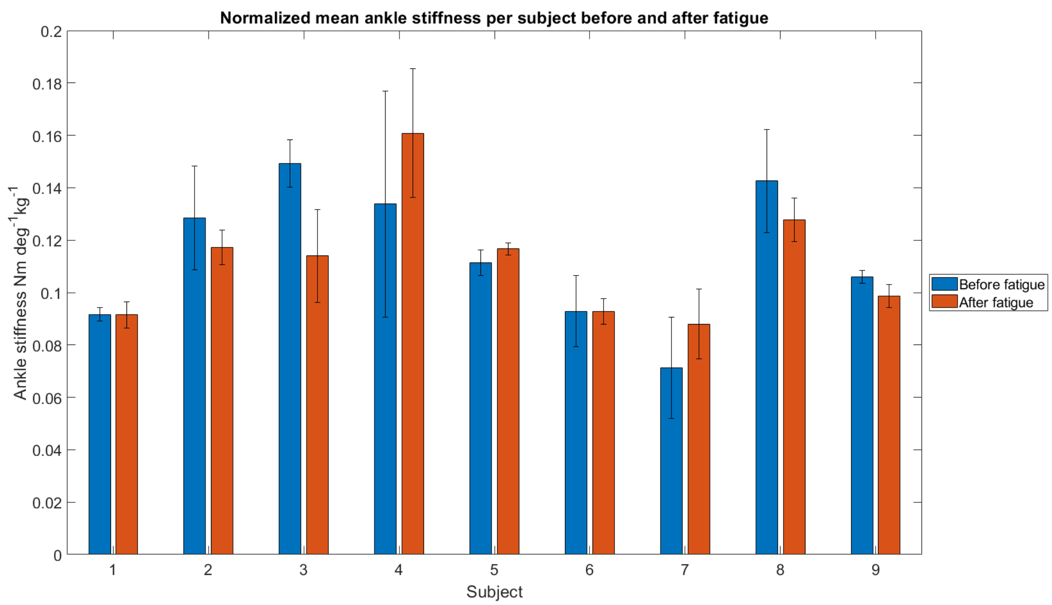

Figure 12: The mean ankle stiffness and its standard deviation for each subject before and after fatigue. The values are normalized by dividing by the subject's mass.

The summed stiffness of the knee and ankle of each subject before and after fatigue is shown in figure 13. The summed stiffness of the knee and ankle did not change significantly when comparing before and after fatigue ($P = 0.2500$). Subjects 4 and 7 show high variability before fatigue, with subject 7 showing high variability after fatigue.

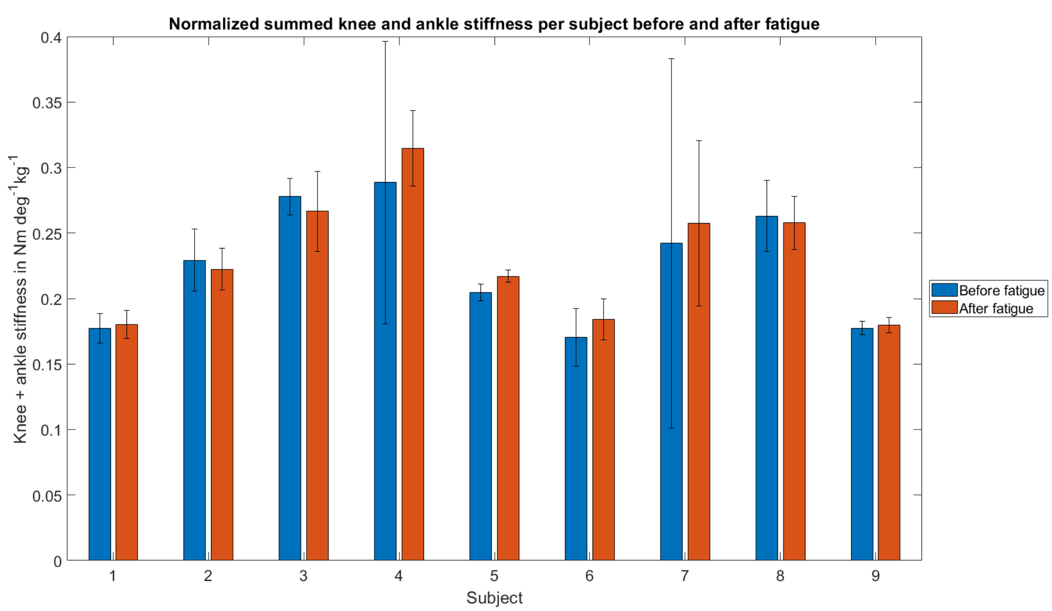

Figure 13: The summed mean knee and ankle stiffness and its standard deviation for each subject before and after fatigue. The values are normalized by dividing by the subject's mass.

The ratio between the knee stiffness and ankle stiffness for each subject before and after fatigue is shown in figure 14. The ratio did not change significantly when comparing before and after fatigue ($P = 0.4258$). Subjects 4, 6 and 7 show high variability before fatigue, with subjects 3 and 7 showing high variability after fatigue.

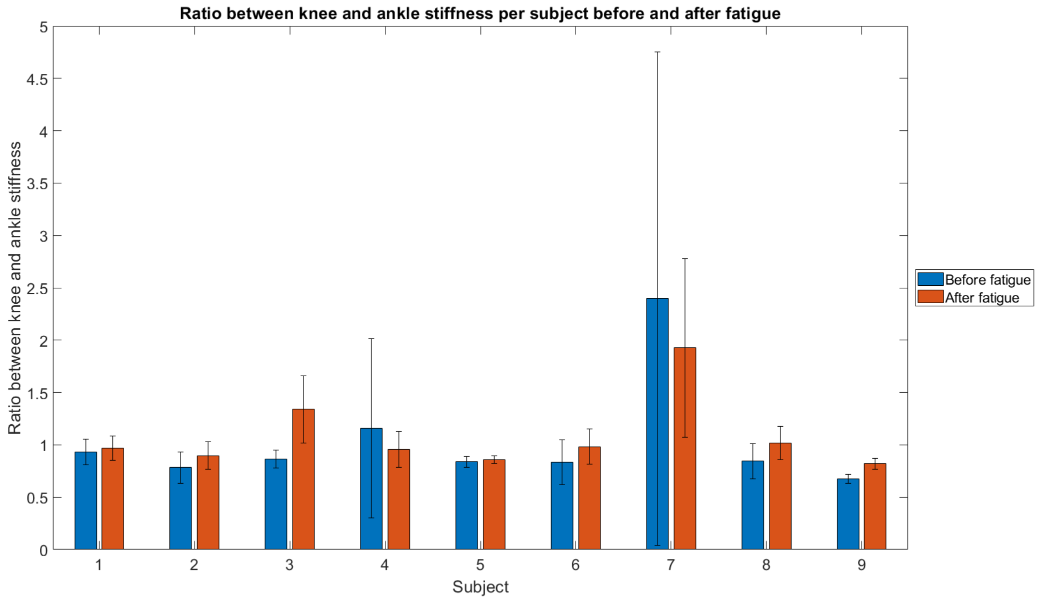

Figure 14: The ratio between the knee and ankle stiffness and its standard deviation for each subject before and after fatigue.

4 Discussion

In this section the results will be interpreted and compared to findings from previous studies. Challenging and limiting factors will be discussed and finally recommendations will be given for future research.

4.1 Interpretation of results

This study aimed at finding the effect of fatigue on the stiffness of the legs and of the hip, knee and ankle joints. It was hypothesized that the leg stiffness would be reduced in the fatigued state and that the ratio between the joint stiffnesses would change as fatigue develops, specifically by an increase of the knee stiffness and a decrease of the ankle stiffness. The hip stiffness was expected to decrease. The results of this study partly support these hypotheses. The leg stiffness was found to be reduced after fatigue, which agrees with the findings of previous studies [22, 25, 35, 39]. It is unknown whether the leg stiffness in this study decreased by a reduction of maximum ground reaction force, an increase of the leg compression or a combination of these factors. Fourchet et al. found both peak vertical ground reaction force and leg compression increased, but leg stiffness decreased as the leg compression increased to a larger extent [22]. Dutto et al. found peak force was maintained and the decrease in leg stiffness was principally caused by an increase of the leg compression [25]. Conversely, Rabita et al. found the leg stiffness decrease was most associated with a decrease of the peak vertical ground reaction force, with leg compression being maintained over the fatiguing run [35]. Although these mentioned studies all found a decrease of the leg stiffness, the cause is different for each one. These different causes can possibly be explained by differences in subject specific characteristics [23]. Therefore, it is not possible to give an exact reason for the decrease of leg stiffness, and deeper analysis of the data would be needed to give an explanation. One example of such deeper analysis could be comparing the leg compression before and after fatigue, and not just the leg stiffness.

The highly variable values of the hip stiffness, being both positive and negative, indicate an issue in the method of determining the hip stiffness. Indeed, figure 9 shows the hip lacks a linear relation between the moment and angle between initial contact and mid-stance. The assumption of a linear relation is the basis on which the determination of the quasi-stiffness of a joint is based. Therefore, the used method for calculating the hip stiffness proved to be unsound. Whether hip stiffness can be determined in another way, will be discussed in section 4.2.1. Although a significant change of the hip stiffness was found when before and after fatigue were compared, this change should be attributed to mere coincidence. Luo et al. was able to determine the hip stiffness, but found no significant hip stiffness-fatigue interactions [27].

A significant change of the knee stiffness was found after fatigue was induced. Specifically, the knee stiffness was found to increase after fatigue. For the other investigated parameters, particularly the ankle stiffness, the summed knee and ankle stiffness and the ratio between the knee and ankle stiffness, no significant changes were found. The increase of the knee stiffness is in line with findings of Weir et al. [26]. That study also found a decrease of the ankle stiffness, which Luo et al. found as well [27]. It is not known why the change in ankle stiffness of this study is not in line with the findings of these two studies. Weir et al. hypothesized that the ankle becomes more compliant as the knee stiffness increases, to maintain overall leg stiffness [26]. It is possible that subject-specific strategies were applied to reduce the leg stiffness even when the knee stiffness was increased. For most subjects this may be a reduction of the ankle stiffness, but that may not necessarily be the case for all subjects. Other possible reasons for the discrepancy between the found ankle stiffnesses are described in section 4.2, where limitations of this study are discussed. Examples of this are a low amount of analyzed trials per subject and the variability of these trials.

Lorimer et al. found the knee and ankle joint stiffness combination to be the most important for assessing changes in leg stiffness for triathletes [20]. That study found that combining joint stiffnesses increased the reliability of the measures, as joint stiffness reliability was poor when assessed individually. The poor reliability of individual joint stiffnesses could also be a potential reason for the high variability and the lack of significant change for the ankle stiffness. However, neither the summed knee and ankle stiffness nor the ratio between these two was found to change significantly. It should be noted that the ratio did not change significantly solely because of subjects 4 and 7, who showed also high variability for their results. The other subjects showed an increase of the ratio between 2.48% and 55.25% as can be seen in table 2. This increase of the ratio is logical, as the knee stiffness was increased for all subjects except subject 4 and 7. Similar to the leg stiffness, deeper analysis of how the knee stiffness increased and ankle stiffness was maintained is recommended. This could show whether the knee stiffness increased by an increase of the net moment on the knee, by decrease of knee range of motion or by a combination of these factors. Similarly, this could show whether ankle moment and range of motion were maintained, or whether a matching increase or decrease of both the ankle moment and range of motion resulted in a maintained ankle stiffness. Different causes for leg- and joint stiffness changes can have different implications for the etiology of injuries. The increase of the knee stiffness when fatigue is induced can significantly increase the odds of sustaining an overuse running injury, as Messier et al. found increased knee stiffness to be significantly related to overuse injuries [44].

The reduction of the leg stiffness when fatigue is induced can imply an increased risk of soft tissue injury, which is associated with decreased leg stiffness [21]. Dutto et al. hypothesise that running while fatigued, with constant speed, may increase tibial accelerations from the decreased leg stiffness, increasing the possibility of injury [25]. This increase of tibial acceleration has been found in multiple studies [13, 29]. Increased tibial accelerations have been found to be related to tibial stress fractures, a common overuse injury [16]. Increased tibial accelerations may also affect shock attenuation, as tibial accelerations are often used in the calculation of shock attenuation [10, 29, 58]. As no accelerations were measured in this study, direct effects on shock attenuation are outside the scope of this study.

4.2 Limitations and recommendations

In this section the limitations influencing the results will be discussed. Recommendations will be given on overcoming these limitations for future research when appropriate.

4.2.1 Hip stiffness

Unfortunately, the used method of calculating the hip stiffness did not seem to be valid. The joint stiffness is, in essence, the slope of a regression line through the joint moment versus joint angle data [21]. As can be seen in figure 9, or any other figure in Appendix A.2, there is no apparent correlation between the hip moment and hip angle in the braking phase. One reason for this is the fact that the range of motion of the hip in the braking phase is small compared to the other joints. Another, more significant reason is the fact that for most subjects the hip angle decreased, increased and decreased again within the braking phase. This in contrast to the knee and ankle, which angles generally only increased in the braking phase. A final reason is the fact that the curve of the hip moments, given by inverse dynamics, were much more variable both inter- and intra-individually compared to the moments of the knee and ankle.

At least two studies were found where the hip stiffness was calculated for running [20, 27]. Unfortunately, their articles do not contain much more information about the method of calculation other than equation 8. The study by Luo et al. [27] specifically mentions the joint moment difference and the joint angle difference to be taken between initial contact and mid-stance, which is identical to the method of this research. It remains unclear where the studies differ and how they were able to determine the hip stiffness between initial contact and mid-stance, as no hip moment versus hip angle graph was shown in their article.

A potential way of determining hip stiffness is by considering the propulsion phase instead of the braking phase. Figure 15 shows the joint moments and angles of the full stance phase. This is the full stance phase of the same subject and fatigue state shown in figure 9. After mid-stance, when the hip flexion angle starts to decrease, a relatively linear relation can be seen between the hip moment and hip angle.

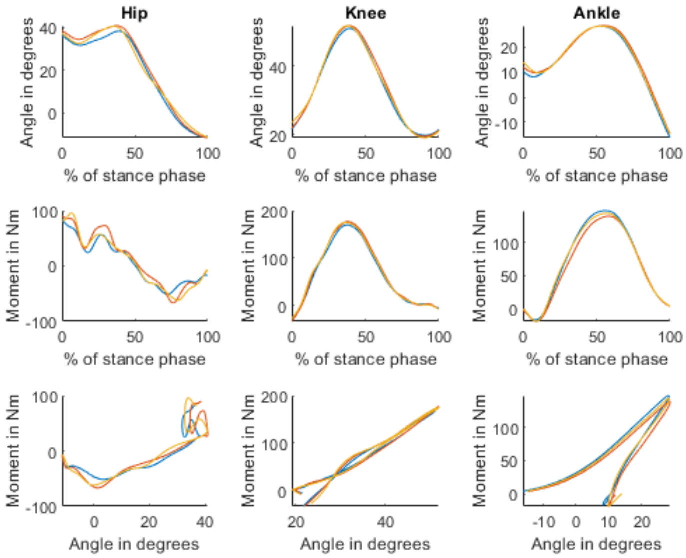

Figure 15: The joint angles and moments between initial contact and toe off of subject 5, before fatigue. The left, middle and right column show the data from the hip, knee and ankle respectively. Top row shows the angle of each joint, while the middle row shows the moment. The bottom row shows the moment plotted against the angle, the linear relationship seen for the knee and ankle is the joint stiffness.

This method is supported by an article by Shamaei et al. [51]. That article found the quasi-stiffness of the hip could be obtained in the resilient loading phase, which consists of terminal stance and initial swing phases. Tateuchi et al. also determined the hip stiffness in the late stance during gait [59]. It should be noted these two studies determined the hip stiffness for walking instead of running, which involves different biomechanics [60]. Still, the relatively linear moment-angle curve between 0° and 30° hip flexion angle in figure 15 (bottom-left graph) would suggest it is possible to determine a hip quasi-stiffness value in the same manner as done by Shamaei et al. and Tateuchi et al.. However, it remains the question what this stiffness value would mean in the context of shock attenuation, as a stiffness determined in the propulsion phase may not necessarily be related to the absorbing and dissipating of shocks. Extra research would be needed to determine whether a relation between shock attenuation and late stance hip stiffness exists.

4.2.2 Data variability

Some subjects, subject 4 and 7 especially, showed a quite high variability of the stiffness values as can be seen in table 2. One primary cause for this is the small number of trials analyzed per subject, as will be discussed in section 4.2.3. Another potential cause can be found in the experiment protocol. Subjects were tasked to run back and forth on a 10-meter runway, with the force plates laying in the middle. In those 10 meters, the subjects had to start running, accelerate up to speed, hit the force plate with their foot, slow down and stop at the end of the runway. These different actions, as opposed to just continuous running, could introduce more variability in the recorded trials. Therefore, it is recommended to utilise a different experimental setup, where the subjects could run continuously while the trials are recorded. One obvious solution would be to use a treadmill, where every step is virtually identical. Riley et al. compared kinematic and kinetic parameters of treadmill running to overground running [61]. It was found that running mechanics on a treadmill are adequately similar to overground running mechanics to be able to generalize treadmill-based running analysis to overground running. Still, researchers might be interested in the biomechanics of overground running specifically, which may disqualify the use of a treadmill. In this instance, it is advised to let the subjects run on a circular or a figure 8 shaped course. This would allow the subject to make more consistent steps, conceivably reducing variability. It should be noted that the straight runway in this study was used partly because of practical reasons, as space in the testing location was limited.

4.2.3 Limited number of trials

Six trials were processed per fatigue state per subject. As only the dominant leg was analyzed, this was reduced to only three trials per fatigue state per subject. This limited number of trials impacts the results as the mean stiffness is more easily skewed from an outlying trial. This fact influences the variability of the results, but also the significance of any findings. The statistical test to determine the significance of changes in stiffness was done based on the subjects' average stiffnesses, and the reliability of these average stiffnesses would be improved with more than three trials to base them on. A treadmill with integrated three-dimensional force plates would again be a good solution. In addition to allowing the subject to run more consistently to give more consistent trials as mentioned in section 4.2.2, it would greatly increase the number of trials for analysis. Practically every step on the treadmill would be a valid trial, which would allow many trials to be recorded in a relatively small time frame. Unfortunately, a treadmill with integrated three-dimensional force plates was not available during this experiment.

Another way to increase the number of available trials would be to use those of the non-dominant leg as well. The non-dominant leg was not included in the analysis of this study as the two legs have been suggested to have distinct mechanical roles during running [62]. One leg is suggested to be propulsive (exerts greater forces) while the other leg is suggested to be supportive (presents smaller flexion-extension action during stance). Results from a study by Pappas et al. seem to agree with this notion, as the dominant leg produced significantly higher force and flight time than the non-dominant leg [63]. However, Brown et al. found no kinematic nor kinetic asymmetries between the legs [64]. That study found both legs to fatigue at a similar rate. Analysis of the non-dominant leg was outside the scope of this study, but could have been interesting to help confirm or deny the existence of leg asymmetries. When no asymmetries would have been found, both legs could have been included in analysis of this study which would increase the number of trials.

5 Conclusion

This study aimed to find the effect of fatigue on the leg and lower extremity joint stiffnesses. For this purpose, the leg, hip, knee and ankle (quasi-)stiffnesses were determined of 9 subjects running before and after fatigue was induced. The leg stiffness was found to decrease after fatigue, while the knee stiffness was found to increase. No linear relation between the hip moment and angle was found between initial contact and mid-stance, which resulted in an inability to determine a hip stiffness. No significant changes were found in the ankle stiffness, nor the ratio between the knee and ankle stiffness nor the summed knee and ankle stiffness. The decrease of the leg stiffness and increase of the knee stiffness can both be associated with overuse injuries, implying running while fatigued increases the odds of sustaining an overuse injury. Deeper analysis of the ground reaction forces, leg compression, joint moments and joint angles is needed to determine the cause of the leg stiffness decrease and knee stiffness increase. It is recommended that future studies utilise a treadmill with integrated force plates to increase the quantity and consistency of trials to analyse.

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A Appendix

A.1 All stiffness values

Table 2: Table with all calculated stiffness values. The stiffness is shown for every subject before and after fatigue, alongside the corresponding standard deviation (ST. DEV.) and coefficent of variation (CV). The CV is calculated using equation 9. The cell containing the CV is colored red when the CV is greater than 20%. Additionally, the percentage difference between the before and after fatigue stiffness is shown. The cell is colored orange when the percentage difference is negative (stiffness after fatigue is lower) and colored blue when it is positive. Note that these values are not normalized to subject mass, in contrast to the bar diagrams in the Results section.

		Leg stiffness			Hip stiffness			Knee stiffness			Ankle stiffness			Knee + Ankle stiffness			Knee / Ankle stiffness (Ratio)		
		k_{Leg}	ST. DEV.	CV	k_{Hip}	ST. DEV.	CV	k_{Knee}	ST. DEV.	CV	k_{Ankle}	ST. DEV.	CV	k_{Knee}+k_{Ankle}	ST. DEV.	CV	k_{Knee}/k_{Ankle}	ST. DEV.	CV
Subject 1	**Before fatigue**	8901	503	5.65%	-4.82	13.91	288.44%	5.13	0.53	10.24%	5.50	0.16	2.88%	10.63	0.68	6.44%	0.93	0.12	13.12%
	After fatigue	8831	350	3.96%	9.73	10.10	103.85%	5.33	0.34	6.42%	5.49	0.30	5.46%	10.82	0.64	5.93%	0.97	0.12	11.88%
	Difference	-0.78%	-	-	-301.74%	-	-	3.80%	-	-	-0.18%	-	-	1.74%	-	-	3.99%	-	-
Subject 2	**Before fatigue**	9598	213	2.22%	11.84	11.82	99.86%	7.54	0.30	3.96%	9.62	1.48	15.41%	17.16	1.78	10.38%	0.78	0.15	19.37%
	After fatigue	8981	527	5.87%	3.08	2.30	74.59%	7.87	0.71	9.05%	8.78	0.49	5.61%	16.65	1.20	7.23%	0.90	0.13	14.66%
	Difference	-6.44%	-	-	-73.94%	-	-	4.36%	-	-	-8.76%	-	-	-3.00%	-	-	14.39%	-	-
Subject 3	**Before fatigue**	12731	177	1.39%	-18.10	13.73	75.88%	10.93	0.42	3.83%	12.67	0.77	6.05%	23.60	1.19	5.02%	0.86	0.09	9.88%
	After fatigue	11812	600	5.08%	-15115.22	26322.70	174.15%	12.96	1.08	8.33%	9.68	1.50	15.54%	22.64	2.58	11.41%	1.34	0.32	23.87%
	Difference	-7.22%	-	-	83424%	-	-	18.61%	-	-	-23.60%	-	-	-4.06%	-	-	55.25%	-	-
Subject 4	**Before fatigue**	10913	1625	14.89%	13.59	30.43	223.97%	12.49	5.23	41.84%	10.78	3.47	32.20%	23.28	8.70	37.37%	1.16	0.86	74.04%
	After fatigue	10302	67	0.65%	-17.55	14.56	82.95%	12.42	0.34	2.75%	12.97	1.98	15.30%	25.38	2.33	9.16%	0.96	0.17	18.06%
	Difference	-5.60%	-	-	-229.20%	-	-	-0.59%	-	-	20.24%	-	-	9.06%	-	-	-17.32%	-	-
Subject 5	**Before fatigue**	8093	26	0.32%	-25.35	7.17	28.28%	7.00	0.12	1.72%	8.33	0.36	4.38%	15.33	0.49	3.17%	0.84	0.05	6.10%
	After fatigue	8125	257	3.16%	-35.80	29.86	83.40%	7.52	0.18	2.43%	8.74	0.17	2.00%	16.26	0.36	2.20%	0.86	0.04	4.43%
	Difference	0.39%	-	-	41.21%	-	-	7.46%	-	-	4.86%	-	-	6.05%	-	-	2.48%	-	-
Subject 6	**Before fatigue**	8389	776	9.25%	44.83	70.87	158.08%	4.93	0.54	11.02%	5.91	0.86	14.58%	10.84	1.40	12.96%	0.83	0.21	25.60%
	After fatigue	8303	699	8.42%	33.50	67.75	202.24%	5.80	0.68	11.73%	5.91	0.31	5.24%	11.71	0.99	8.46%	0.98	0.17	16.97%
	Difference	-1.02%	-	-	-25.28%	-	-	17.74%	-	-	-0.05%	-	-	8.04%	-	-	17.80%	-	-
Subject 7	**Before fatigue**	10876	313	2.88%	10.01	2.71	27.08%	13.56	9.65	71.14%	5.66	1.54	27.22%	19.22	11.19	58.21%	2.40	2.36	98.36%
	After fatigue	10854	649	5.98%	7.71	5.90	76.51%	13.47	3.97	29.44%	6.99	1.05	15.02%	20.46	5.01	24.52%	1.93	0.86	44.46%
	Difference	-0.21%	-	-	-23.02%	-	-	-0.67%	-	-	23.46%	-	-	6.44%	-	-	-19.54%	-	-
Subject 8	**Before fatigue**	9758	98	1.01%	-13.50	12.18	90.16%	8.35	0.51	6.10%	9.88	1.36	13.78%	18.23	1.87	10.26%	0.85	0.17	19.88%
	After fatigue	9276	32	0.35%	-223.64	137.55	61.50%	9.01	0.82	9.15%	8.85	0.58	6.50%	17.87	1.40	7.83%	1.02	0.16	15.64%
	Difference	-4.94%	-	-	1556%	-	-	7.93%	-	-	-10.38%	-	-	-1.99%	-	-	20.43%	-	-
Subject 9	**Before fatigue**	7706	638	8.27%	-1.32	2.23	169.34%	3.92	0.16	4.07%	5.82	0.13	2.25%	9.74	0.29	2.98%	0.67	0.04	6.32%
	After fatigue	7382	536	7.26%	-16.63	24.48	147.26%	4.44	0.07	1.55%	5.42	0.24	4.47%	9.86	0.31	3.15%	0.82	0.05	6.01%
	Difference	-4.21%	-	-	1163%	-	-	13.21%	-	-	-6.82%	-	-	1.25%	-	-	21.49%	-	-

A.2 Joint angles and moments

Figure 16-33 show the joint angles and moments of all subjects between initial contact and mid-stance.

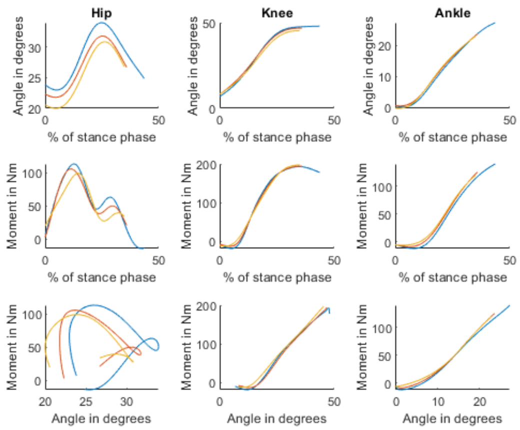

Figure 16: Joint angles and moments of subject 1, before fatigue.

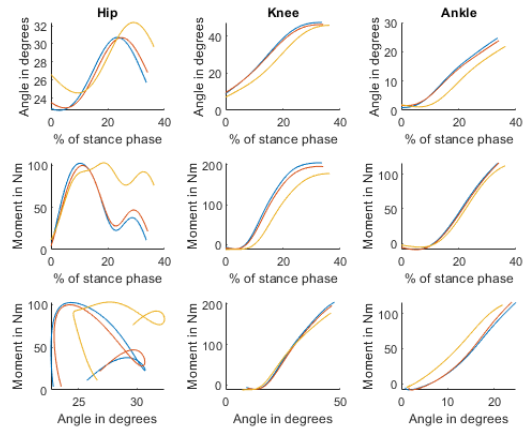

Figure 17: Joint angles and moments of subject 1, after fatigue.

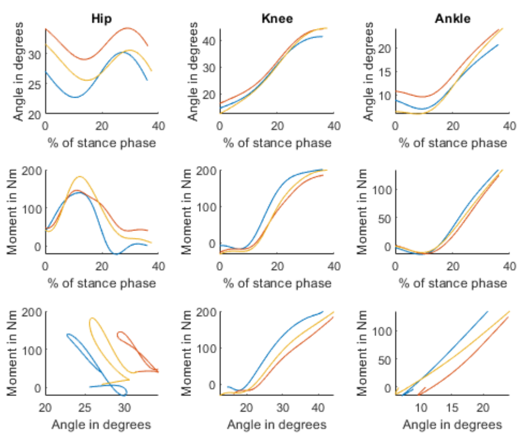

Figure 18: Joint angles and moments of subject 2, before fatigue.

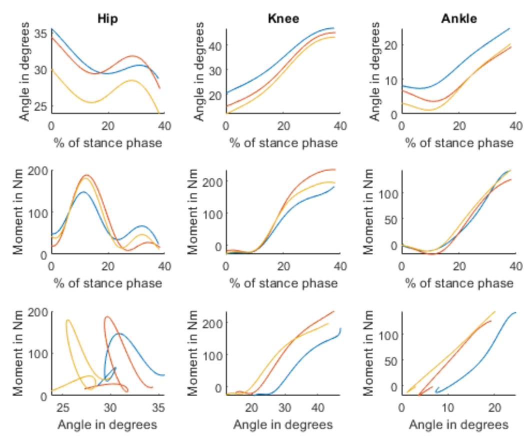

Figure 19: Joint angles and moments of subject 2, after fatigue.

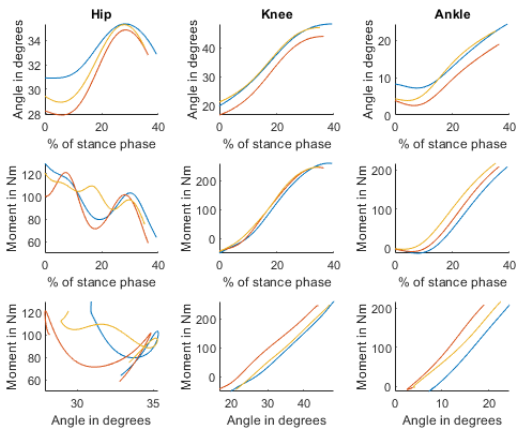

Figure 20: Joint angles and moments of subject 3, before fatigue.

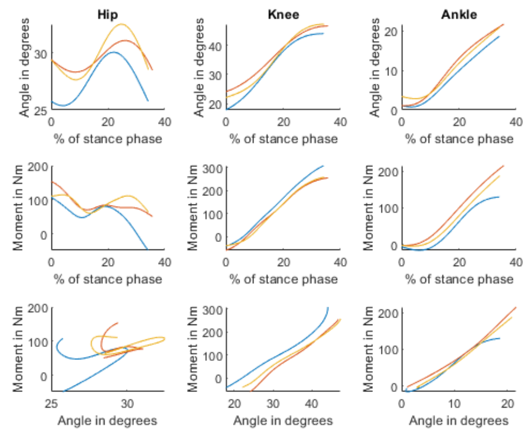

Figure 21: Joint angles and moments of subject 3, after fatigue.

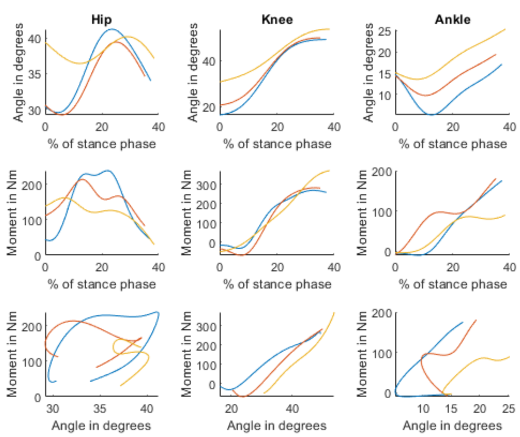

Figure 22: Joint angles and moments of subject 4, before fatigue.

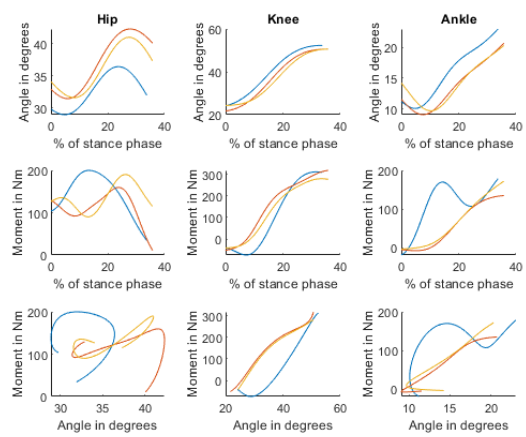

Figure 23: Joint angles and moments of subject 4, after fatigue.

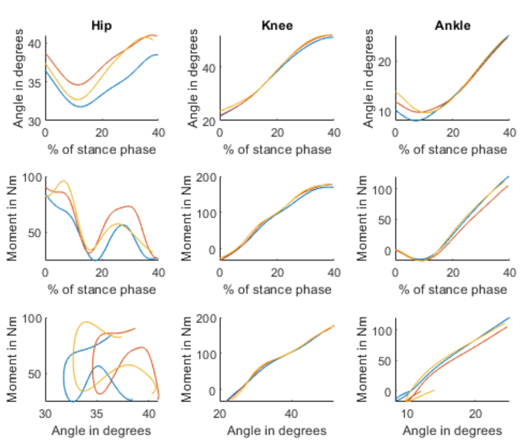

Figure 24: Joint angles and moments of subject 5, before fatigue.

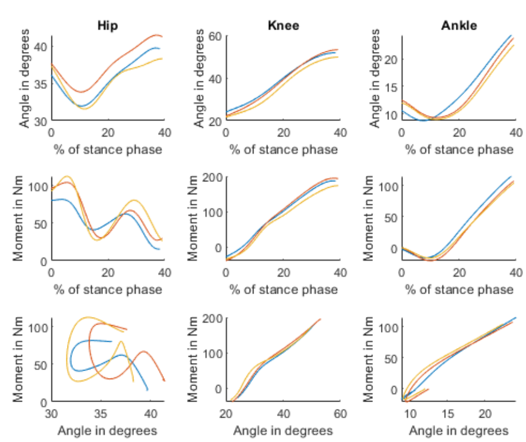

Figure 25: Joint angles and moments of subject 5, after fatigue.

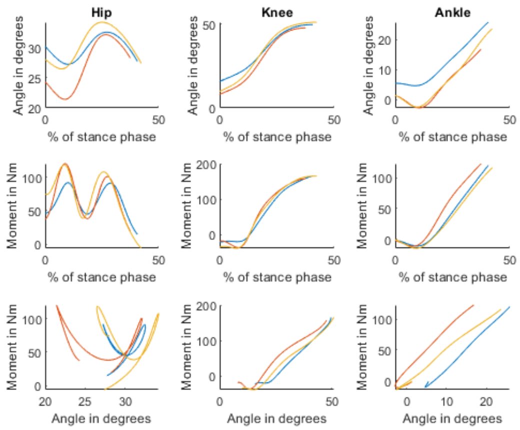

Figure 26: Joint angles and moments of subject 6, before fatigue.

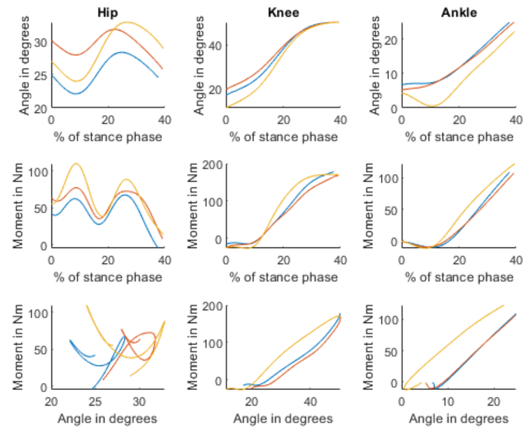

Figure 27: Joint angles and moments of subject 6, after fatigue.

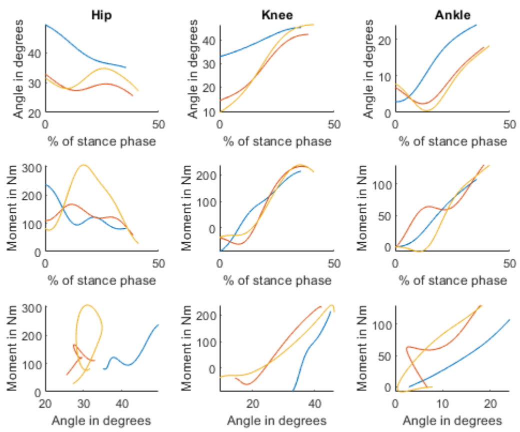

Figure 28: Joint angles and moments of subject 7, before fatigue.

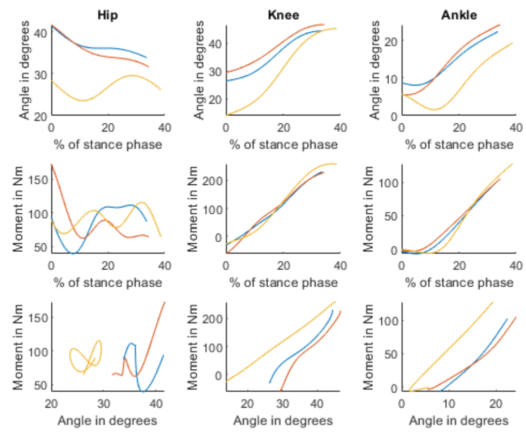

Figure 29: Joint angles and moments of subject 7, after fatigue.

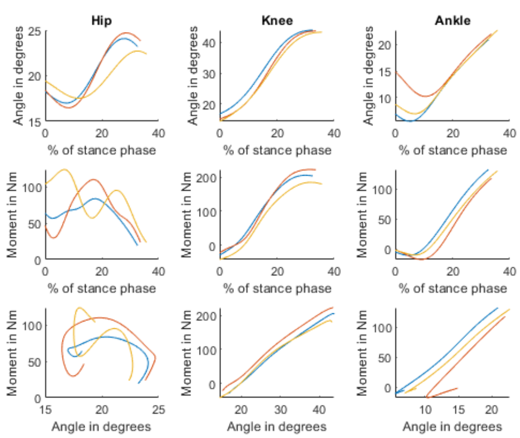

Figure 30: Joint angles and moments of subject 8, before fatigue.

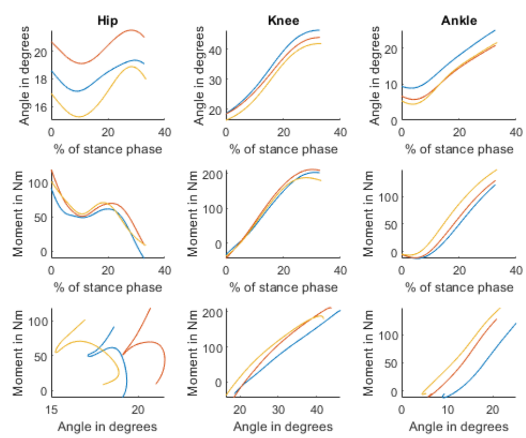

Figure 31: Joint angles and moments of subject 8, after fatigue.

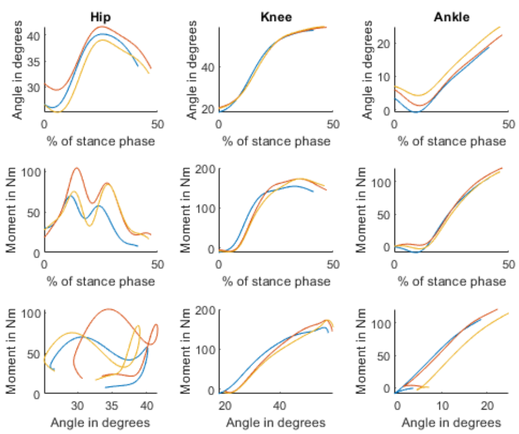

Figure 32: Joint angles and moments of subject 9, before fatigue.

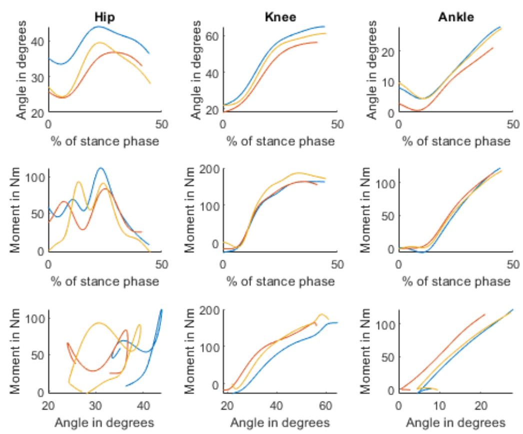

Figure 33: Joint angles and moments of subject 9, after fatigue.