COMBINED EFFECTS OF FOOT PLACEMENT TECHNIQUE AND SPORT SPECIFIC HABITUAL TRAINING ON LANDING MECHANICS

by

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ACADEMIC ETHICS AND INTEGRITY STATEMENT

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ABSTRACT

COMBINED EFFECTS OF FOOT PLACEMENT TECHNIQUE AND SPORT SPECIFIC HABITUAL TRAINING ON LANDING MECHANICS

Correct technique that provides efficient and safe dissipation of ground reaction forces (GRF) is crucial during landing motions. Parkour practitioners (traceurs) intrinsically developed a landing technique in which they solely land on their forefoot (FFL) to counter extreme loading demands of their practice. Traceurs attenuate impact forces substantially during FFL compared to traditionally used toe-heel landing (THL). However, traceurs are already expected to execute their habitual foot placement technique (FPT), FFL, more efficiently via favorable adaptations in their musculoskeletal and neuromuscular system. Up till now, it has not been tested specifically if FFL is superior over THL in shock absorption, regardless of performers habitual and/or preferred FPT. Additionally, it is not known how sport specific habitual training (SSHT) in FPT affects landing mechanics. Presently, to fill those gaps FFL and THL mechanics were analyzed during drop landings from 75 cm in three groups (n = 3 x 12); (1) traceurs (habitual FFL practice), (2) basketball players (habitual THL practice) and (3) non-athletes (no habitual training). GRF metrics, lower body joint kinematics, activation patterns of tibialis anterior (TA) and gastrocnemius medialis (GM) muscles, and mechanical characteristics of the Achilles tendon (AT), TA and GM were measured to examine the effects of FPT and SSHT on landing mechanics. Results indicate that FFL is considerably advantageous than THL for shock absorption, independently of SSHT and preferred FPT. Furthermore, kinematic and neuromuscular strategies were altered between groups to achieve similar shock attenuation for each technique, based on the potential adaptive effects of their SSHT. These findings present valuable insights into the effects of FPT on landing mechanics which may have practical implications for every practitioner and trainer whether in sports or recreational activities.

Keywords: Parkour, Forefoot, Ankle, GRF, Kinematic, EMG, Stiffness, Recoil.

ÖZET

AYAK YERLEŞTİRME TEKNİĞİ VE SPORA ÖZGÜ HABİTUEL ANTRENMANIN İNİŞ MEKANİĞİ ÜZERİNDEKİ BİRLEŞİK ETKİLERİ

Doğru teknik, iniş hareketleri sırasında yer tepki kuvvetlerinin (YTK) verimli ve güvenli bir şekilde sönümlenmesini sağlar. Parkur sporcuları (parkurist) branşlarının gerektirdiği yüksek yükleri karşılayabilmek için ön ayak üzerine indikleri bir teknik (ÖAİ) geliştirmişlerdir. Parkuristler ÖAİ ile geleneksel parmakucu-topuk inişe (PTİ) oranla darbeyi önemli derecede daha fazla sönümlerler. Fakat, parkuristlerin kas-iskelet ve nöromusküler sistemlerindeki adaptasyonlarından dolayı zaten habituel ayak yerleştirme teknikleri (AYT) olan ÖAİ'i daha etkin uygulamaları beklenmektedir. Şu ana kadar, habituel ve/veya tercih edilen AYT'nden bağımsız olarak ÖAİ'in PTİ'den üstünlüğü ve spora özgü habituel antrenmanın (SÖHA) iniş mekaniğini nasıl etkilediği test edilmemiştir. Bu açıklıkları doldurmak için 75 cm'den yapılan iniş hareketi sırasında ÖAİ ve PTİ mekanikleri üç ayrı grupta incelenmiştir (n = 3 x 12): (1) Parkuristler (ÖAİ içeren habituel antrenman), (2) basketbolcular (PTİ içeren habituel antrenman) and (3) sporcu olmayanlar (habituel antrenman yok). AYT ve SÖHA'ın iniş mekaniği üzerindeki etkilerini incelemek üzere YTK, alt vücut eklem kinematiği, tibialis anterior (TA) ve gastrocnemius medialis (GM) kaslarının aktivasyon paternleri, ve aşil tendonunun, TA ve GM kaslarının mekanik karakteristikleri ölçülmüştür. Sonuçlar göstermiştir ki, SÖHA'a ve tercih edilen AYT'ne bağlı olmaksızın ÖAİ tekniği PTİ tekniğinden şok absorbsiyonu açısından önemli ölçüde daha avantajlıdır. Fakat gruplar, her teknik için benzer şok emilimi değerlerini sağlamak için, SÖHA'ın yarattığı olası adaptatif etkilerden kaynaklandığı düşünülen farklı kinematik ve nöromusküler stratejiler uygulamışlardır. Bu bulgular, sporda veya rekreasyonel aktivitelerde hem uygulayıcılar hem de antrenörler için pratik uygulanımları olabilecek, AYT'nin ve SÖHA'ın iniş mekaniği üzerindeki etkileri hakkında değerli bilgiler sunmaktadır.

Anahtar Sözcükler: Parkur, Ön Ayak, Ayak Bileği, YTK, Kinematik, EMG.

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LIST OF SYMBOLS

m	Mass
W	Weight
a	Acceleration
g	Gravity
F	Force

LIST OF ABBREVIATIONS

GRF	Ground Reaction Forces
COM	Center of Mass
BW	Body Weight
ACL	Anterior Cruciate Ligament
pVF	Peak Vertical Forcel
IC	Initial Contact
LR	Loading Rate
EMG	Electromyography
sEMG	Surface Electromyography
CNS	Central Nervous System
SLRA	Short Latency Reflex Activity
LLRA	Long Latency Reflex Activity
GM	Gastrocnemius Medialis
TA	Tibialis Anterior
AT	Achilles Tendon
COM	Center of Pressure
THL	Toe-Heel Landing
HTL	Heel-Toe Landing
FFL	Forefoot Landing
FP	Force Platform
BW	Body Weight
ROM	Range of Motion
MVIC	Maximum Voluntary Isometric Contraction
ANOVA	Analysis of Variance
SSC	Stretch Shortening Cycle

1. INTRODUCTION

In daily life; humans, as well as all beings on earth, are constantly affected by the gravity and thus the gravitational forces. These forces not only act upon the external structures of the human body, all internal structures are also under a steady pull because of gravitational forces. In addition, every time a person contacts the ground or any other surface, reaction forces operate on the body, in the opposite direction to gravitational forces, according to Newton's second and third law. By far the most frequent and hence notable one of the reaction forces is the ground reaction forces (GRF), which represents forces exerted by the ground on the body through the center of mass (COM) (Figure 1.1).

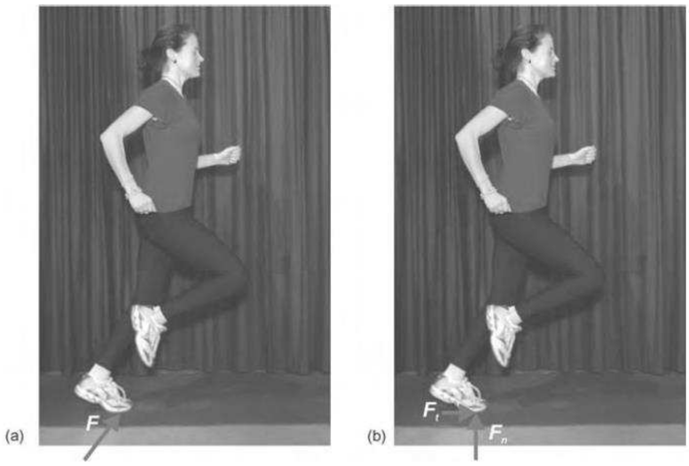

Figure 1.1 (a) GRF and (b) its components [1].

GRF act upon the body in the presence of any ground contact, whether in a static or dynamic condition. However, the magnitude of GRF increases from static to dynamic conditions, taken the mass (m) is constant [2]. During a static condition such as standing the GRF equals to weight (W = mg) of the person. If the person aims to move, he/she has to exert additional forces on the ground by activating his/her muscles. According to Newton's second law, the additional force causes acceleration. The magnitude of acceleration depends of the person's mass (F = ma). Eventually, the total GRF should be higher than person's weight for movement to happen.

Furthermore, as the speed of movement increases, the impact during the ground contact and hence GRF increases [3–6]. A person needs to push the ground more forcefully to accelerate or decelerate the body for faster or slower movements. During locomotive movements such as walking, jogging and running; peak vertical GRF are about 1.42, 2.35 and 2.46 times of body weight (BW) at the speeds of 2 m/s, 4 m/s and 5 m/s respectively [4]. During locomotion, the aim is to transfer COM forward, for this reason at faster movements speeds over vertical GRF tends to decline or plateau [4, 6]. As for more rigorous movements such as jumping and landings, the GRF could reach tremendous amounts [7–10]. It has been reported that during drop landing, GRF can reach up to 4.8, 6.0, 7.8 and 11.0 BW during landings from 30, 60, 90 and 128 cm, respectively [11, 12]. Note however that, between jumping (acceleration) and landing (deceleration), the landing movement is the riskier one, injury wise, because of the stored kinetic energy of the body before ground contact. Unlike the acceleration phase of a jump, during landing the kinetic energy is not introduced to the body by controlled voluntary contractions of muscles. The body has to absorb the kinetic energy and disperse resultant GRF safely to stop the motion during landing. However, this task may exceed the capabilities of the body if the musculoskeletal system is not ready.

Following the ground contact, the contact forces are distributed through bones, joints, connective tissues and muscles. Therefore, these forces may put high amounts of stress on the musculoskeletal system, especially during rigorous movements [13]. If the stress on the tissues are not absorbed and dissipated effectively, injuries most likely happen [14, 15]. These injuries may occur in the form of acute or chronic traumas.

On the other hand, overuse injuries such as stress fractures, tendinosis and ligament tears are pretty common in physically active people, particularly in athletes [16–20]. Repetitive loading of these tissues results in micro traumas. Without proper rest to let the tissue heal and adapt, injuries may occur eventually. Additionally, poor landing technique further elevates the injury occurrence [21–23].

In this context, correct technique and controlled deceleration is crucial to safely absorb and disperse GRF throughout the body during landing, particularly for the athletes who frequently perform landings in their practice. The efficiency in shock absorption is accomplished by optimizing landing mechanics.

1.1 Landing Mechanics

The term landing mechanics represents the kinetics, kinematics and neuromuscular mechanisms of musculoskeletal system during landing motions. These issues are addressed below.

1.1.1 Landing Kinetics

Landing kinetics mainly examines the GRF and related metrics such as peak force, loading rate, rate of force development, pressure distribution and center of pressure. Nevertheless, forces acting on internal structures of musculoskeletal system such as joints, tendons, ligaments and bones, as well as muscle forces can also be analyzed by inverse dynamics methods [2].

Force platforms are the most common equipment for measuring GRF. There are many kinds of force platforms such as strain gauge, piezoelectric and piezoresistive ones. Even though the main principle in measuring the force remains the same, characteristics and outputs may differ among various force platforms, which can present advantages or disadvantages depending on the study aims. The sampling rate (frequency response) and the dimension of force vector are the most notable distinctions. For example, piezoresistive pressure sensors present sensitive information such as pressure distribution of contact area, however sampling rate may be low and more importantly the force vector may be obtained only for the vertical axis. In contrast, strain gauge and piezoelectric sensors present forces in 3-D and for high sampling rates, but lack the capacity of measuring pressure distribution [2].

As a 3-D vector, the GRF is examined in three axes; vertical (longitudinal), anteroposterior (sagittal) and mediolateral (frontal) (Figure 1.2).

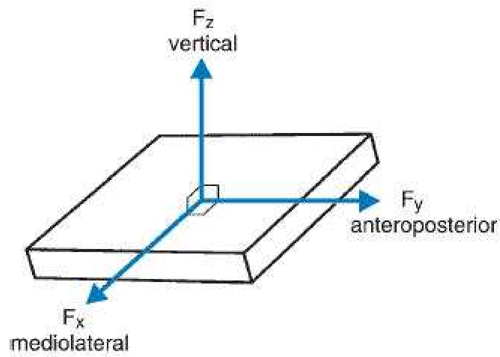

Figure 1.2 Representation of 3-D GRF vectors on a force platform [24].

The anterioposterior and mediolateral GRF act as reactional forces to the ground friction. Those forces are more apparent during forward or lateral propulsion such as sprinting, long jump and cutting motions. However, the vertical GRF is considerably higher than sagittal and frontal GRF during landing, especially for landings with minimum anterioposterior or mediolateral motion such as drop landings from a platform [25]. Therefore, the following metrics will be defined for the vertical GRF.

Peak vertical GRF, also referred briefly as the peak vertical force (pVF), defines the maximum vertical force value during the motion i.e., in the present case, landing (Figure 1.3). In addition to pVF, the following metrics are commonly examined; (i) time from initial contact (IC) with the ground to the point of pVF (time to pVF) and (ii) the rate of force development from IC to pVF, defined as loading rate (LR) (Eq. 1.1). The ability of the musculoskeletal system to respond to the pVF, not only depends on the magnitude of the force but also the time interval between IC and pVF [26].

$$LR = \frac{\text{pVF}}{\text{time to pVF}} \tag{1.1}$$

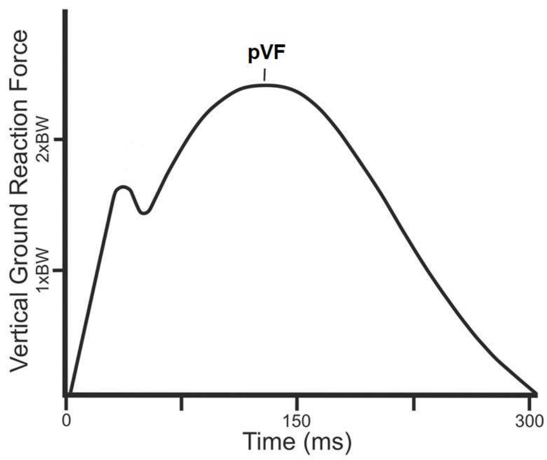

Figure 1.3 Representation of the point of pVF on force-time graph [27].

1.1.2 Landing Kinematics

Kinematic evaluation of landing requires acquisition of two or three-dimensional trajectories of selected body parts or the whole body. Linear or angular displacement as well as velocity and acceleration profiles can be derived after data extraction (Figure 1.4). The most commonly used equipment for motion analysis is video camera, nevertheless other systems such as inertial sensors are also used.

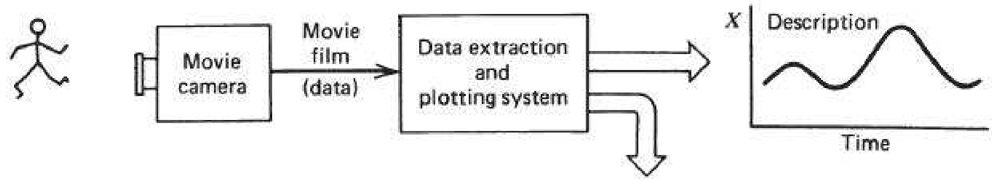

Figure 1.4 Data acqusition and analysis process of motion analysis [28].

The behavior of musculoskeletal system is very complex during dynamic movements. If all the factors and relative parameters are taken into account, it is extremely hard to analyze and draw a conclusion from body movements. It is therefore, required to incorporate convenient models for kinematic analyses. Consequently, assumptions are made to simplify the analysis process. The most common one in kinematics is the 'rigid body' assumption, in which, body parts are assumed to be undeformable. It should be noted that rigid body mechanics are also used in kinetics to analyze the external and internal forces on different interconnected parts of the body. In landing kinematics, mainly the joint angles, angular velocities and acceleration are examined in a link-segment model according to rigid body mechanics (Figure 1.5). For this, joint centers must be derived from the segments.

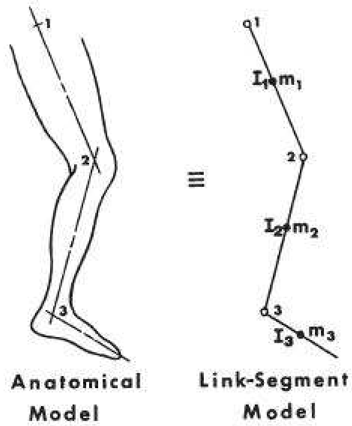

Figure 1.5 Transition from anatomical model to link-segment model using rigid body [28].

Marker placements are made according to the link-segment model for motion analysis. The coordinates of these markers in 2-D or 3-D plane need to be acquired to extract and plot at regular intervals in time for further analysis (Figure 1.6). One camera may be enough for a 2-D motion analysis, however at least 2 cameras are needed for a 3-D motion in theory. Even so, far more than 2 cameras should be used in order to capture marker positions from different angles in every frame during dynamic measurements. It should be noted that in a 2-D motion analysis, bilateral symmetry is assumed in the sagittal plane. In addition, joint movements in the frontal and transverse plane may create errors, because those movements cannot be captured. Therefore, extra care must be taken to minimize these errors during data acquisition in 2-D motion analysis.

Landing mechanics is mostly interested in the kinematics of lower extremity, including the hip joint. Therefore, foot, shank, thigh and torso (if needed pelvis and arm) are defined as rigid bodies. Anatomical marker placements are to be made according to this. There are numerous protocols for anatomical marker placement [29, 30]. The

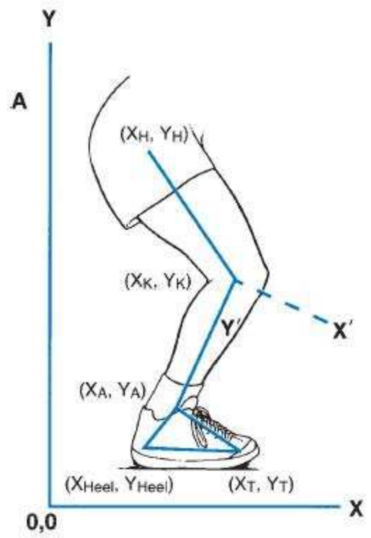

Figure 1.6 Representation of 2D coordinate system on sagittal plane [24].

suitable protocol must be selected according to the analyzed motion, equipment and objective of the study.

1.1.3 Neuromuscular Mechanisms in Landing

Muscles contract voluntarily or reflexively to produce force and decelerate movement during landings. The neuromuscular system acts to control these muscular contractions and, hence, joint moments by excitatory or inhibitory mechanisms. The electrical signal that facilitates muscle contraction is measured by electromyography (EMG). EMG does not directly measure force outputs of muscles, but it gives information about force contribution of the muscles [31].

Dynamic movements are mostly measured by surface EMG (sEMG), by an electrode placed on the surface (skin) of a muscle. sEMG measures and records the sum of motor unit action potentials conveyed along the muscle fibers [26]. sEMG measurements can be affected by intrinsic or extrinsic factors such as electrode placement site (detection location), orientation of the electrode, skin resistance, and signal artifacts and noise [31]. Therefore, minimizing and standardizing these effects are crucial to acquire valid and reliable sEMG recordings.

Muscular contractions before and during (post-activation) landing motion are controlled by the central nervous system (CNS) and peripheral feedback [26,32] (Figure 1.7). The muscles of lower extremity start to activate (pre-activation) before foot contact to prepare for the GRF [33]. Pre-activation is programmed and modulated by the CNS through sensory input (visual, vestibular and somatosensory) and predicted impact time and force [34, 35]. After initial contact, hip and knee extensors and ankle plantar flexors are forced to stretch which leads to stretch reflex responses to occur. The magnitude of stretch reflex responses not only depends on velocity of stretch but also excitation level of the motor neuron pool at the time of stretch, which is related to pre-activation level. Pre-activation increases muscle spindle sensitivity by co-activation of alpha and gamma motor neurons [35]. Therefore, without pre-activation, stretch reflex responses would occur too late to safely absorb the impact by modulating joint rotations.

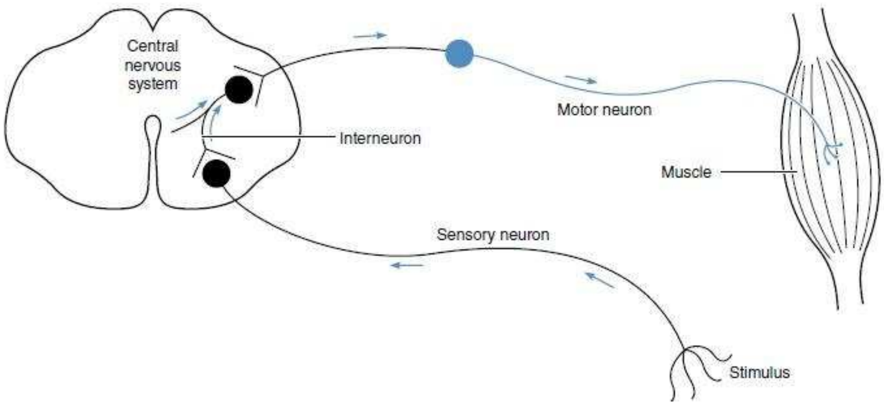

Figure 1.7 Schematic diagram of motor control of muscle activation by CNS and sensory feedback [36].

Short latency reflex activity (SLRA) of the gastrocnemius medialis (GM) and soleus muscles peak at around 53.2 and 55.7 ms after landing from 0.45 m, respectively [26]. Considering that the peak GRF occurs between 30-45 ms after IC [37, 38], solely SLRA would not be enough to efficiently control and decelerate fast ankle joint rotations without pre-activation and continuing central motor control after IC [32]. Long latency reflex activity (LLRA) occurs along with voluntary activation after 80 ms for GM from 0.30 m [39, 40]. Despite that, studies are limited on LLRA during landings.

Co-activation of the joint musculature is also crucial for a safe and efficient energy dissipation by modulating joint stiffness during landing [41]. Modulation of stiffness by co-contraction starts from pre-landing and continues throughout landing [8,35]. The stiffness regulation is necessary because optimum stiffness differs according to landing surface (hard or soft), height (because of varying joint rotation velocities), available feedback information, technique and age [34, 42–47]. The tibialis anterior (TA) co-contracts with the triceps surae muscles, as the main dorsi-flexor of the ankle joint during landing. Activity of the TA starts as early as 108 ms along with GM (from 0.80 m) before IC [33], even though pre-activation of plantar-flexors are higher [8], and increases post IC, peaking at around 26.5 ms (from 0.45 m) [26]. However, other studies reported later peak times for TA [39, 48].

1.2 Landing in Sports

Rigorous landing movements are frequently performed in many athletic activities such as basketball, volleyball, gymnastics and handball. The loads and the resultant GRF can reach tremendous amounts during landings in sports, in order to enhance athletic performance [8, 12]. For this reason, efficient and safe dissipation of GRF is even more crucial for athletic movements.

Athletes developed landing techniques and strategies to compensate for these demands and increase their performance, particular to the requirements of their branch of sports [8,49,50]. Correct jumping technique may be crucial for a safe landing [51,52]. However, it should be noted that safety and performance may not go hand in hand every time in competitive sports [53,54]. Sometimes athletes may risk injury to increase their sportive performance. In any case, athletes perform their preferred landing techniques repetitively during their sports practice. In time, repetitive practice of these techniques results in favorable adaptations in musculoskeletal and neuromuscular system according to the principle of adaptation, which in return may further increase the efficiency of practiced technique [8, 55–64].

Among all of these sports, parkour stands out the most in regards to landing movement, because of its unique loading demands and sports specific landing techniques.

1.2.1 Landing in Parkour

Parkour is an urban sport that was adopted from military training and developed to be practiced in the suburbs. The core principle is to reach from one point to another by overcoming obstacles in the fastest and the most efficient way by only means of parkour practioner's (traceur) own body. Running, jumping, landing, climbing, vaulting and rolling movements are widely used in parkour. However, movement possibilities are vast, being only limited by the traceur's imagination and physical capacity.

Jumping and landing movements, mostly from high obstacles or structures, make up a significant portion of parkour practice (Figure 1.8). As exemplified before, the vertical GRF is directly proportional to the height from which the landing is executed. Differently from any other sports that require landing from extreme heights, such as gymnastics and olympic high jumps, traceurs mostly land on hard surfaces. Landing on a hard surface can further increase the vertical GRF and LR [65–67]. Therefore, landings in parkour may put exceptional loads on the body unlike in any other sports practice. This creates the necessity for a correct technique and controlled deceleration during parkour landings to prevent injury occurrence.

Traceurs developed a unique landing technique, named precision landing. In precision landing, the traceur lands only on his/her forefoot not letting his/her heel or rear-foot touch the ground. The precision landing technique enables traceurs to transfer their COM forward for fluent subsequent movement and also land on narrow surfaces such as rails and thin walls, which are common obstacles in parkour (Figure 1.9).

Figure 1.8 Parkour landing from a high structure [68].

The most distinct characteristic of precision landing is foot placement. During landings in recreational activities and almost all of the sports, all plantar surface of the foot contacts the ground; either flatfoot, toe then heel (THL) or heel then toe (HTL) [22, 70–72]. In precision landing only the forefoot touches the ground in plantar flexed position without any subsequent heel contact. Landing with only on forefoot is defined as forefoot landing (FFL). Other than traceurs, only ballet dancers perform FFL in their practice [73]. However, kinematics of ankle, knee and hip joints widely differ between parkour and ballet practice because of distinct performance requirements.

FFL mechanics has been examined in previous studies [9, 10, 70]. Gross and Nelson reported 22% less vertical GRF with FFL compared to THL during landing from a countermovement jump in male basketball players [9]. Kovacs et al. compared FFL and HTL during drop jump from 0.4 m in male non-athletes [70]. Average pVF was found to be lower by 39.1%, and time to pVF higher by 53 ms in FFL compared to HTL. There were also significant differences in angular displacement of the hip, knee and ankle during flexion phase. Hip and knee angular displacement was higher for HTL, while ankle displacement was higher for FFL. The EMG of GM was reported to be significantly higher for FFL on average by 32.7%. Self and Paine examined FFL with stiff knee from a height of 0.30 m and compared GRF and Achilles tendon (AT) forces with natural, natural foot placement + stiff knee and flat foot + stiff knee

Figure 1.9 Parkour precision landing on a rail [69].

landings [10]. The pVF was achieved by FFL with stiff knee, while AT forces were reported to be highest for FFL.

Additionally, FFL mechanics in parkour precision landing have been studied in the recent years [50,72,74,75]. Puddle and Maulder reported significantly less pVF (by 38.4%) and LR (by 54.2%), and slower time to pVF (by 60.6%) with precision landing from 0.75 m with forefoot placement compared to traditional one with toe-heel placement in traceurs [50]. In their following study, they compared traceurs' precision landing performance with recreationally trained individuals' preferred landing technique, which is reported to be THL by 91.7% (8.3% forefoot to mid-foot) [72]. FFL in traceurs elicited significantly less pVF, LR and slower time to pVF than THL in recreationally trained. Maldonado et al. compared postural control between traceurs' FFL practice and non-athletes' preferred landing technique from 0.3 and 0.6 m [76]. Traceurs showed better postural control performance with FFL compared to non-athletes' performance with their preferred landing technique. Additionally, FFL resulted in a more anteriorly positioned center of pressure (COP) from IC to post 800 ms. In their most recent study, Maldonado et al. examined dynamic and kinematic parameters [75]. They compared traceurs' FFL performance and non-athletes' landing performance with their preferred technique from 0.3 and 0.6 m. Higher hip, knee and ankle joint ROM, and lower pVF values were reported for FFL performance of traceurs. The participants performed their preferred and/or habitual landing techniques in all of these studies, which may affect landing performance.

Repetitive practice of movement and loading patterns may elicit specific adaptations in musculoskeletal and neuromuscular system [55, 64, 77–82]. Grospretre and Lepers showed that traceurs' knee extension torques are higher than gymnasts, power athletes and control participants [58]. This may be the effect of parkour practice, which includes frequent and forceful eccentric muscle actions during landings and plyometric movements, as discussed by those authors. Eccentric effort of knee extensors is a significantly important part of deceleration during landings, as often incorporated by the traceurs in their practice [83]. Also, muscle and tendon mechanical characteristics of traceurs may be adapted to high impact forefoot landings because of high habitual loading and repetitive FFL practice.

1.3 Aim of the Study

Based on the background provided, traceurs are expected to execute their habitual foot placement technique, FFL, more efficiently via favorable adaptations in their musculoskeletal and neuromuscular system, compared to other groups that do not perform FFL. Similarly, athletic groups that frequently perform THL in their practice may be expected to be more efficient in THL compared to traceurs. However, it has not been tested specifically if FFL is superior over THL in shock absorption regardless of performer's habitual and/or preferred foot placement technique. Additionally, it is not known how sport specific training in foot placement technique affects landing mechanics.

Presently, the aim is to fill those gaps by examining the effects of habitual and/or preferred foot placement technique on landing mechanics. 3 groups were included in the study for this purpose: (1) Traceurs who frequently perform FFL during their practice. (2) Basketball players who frequently perform THL during their practice. (3)

Non-athletes who do not regularly practice either of these techniques. In addition to GRF metrics, we analyzed lower body joint kinematics (hip, knee and ankle), activation patterns of tibialis anterior (TA) and gastrocnemius medialis (GM) muscles, and mechanical characteristics of the Achilles tendon (AT), TA and GM to further observe the effects of technique and habitual training, and interaction between those factors.

2. METHODS

2.1 Participants

Experimental procedures were in strict agreement with the guidelines and regulations concerning human welfare and experimentation set by Turkish law and approved by a Committee on Ethics of Human Experimentation at Bogazici University, Istanbul.

Three groups each of which include 12 male traceurs (age: 18.6 $\pm$ 2.1 years; body mass: 64.7 $\pm$ 7.2 kg; and height: 1.73 $\pm$ 0.06 m), basketball players (age: 19.8 $\pm$ 2.2 years; body mass: 81.2 $\pm$ 9.3 kg; and height: 1.85 $\pm$ 0.08 m) and, non-athletes (age: 19.0 $\pm$ 1.9 years; body mass: 66.0 $\pm$ 6.5 kg; and height: 1.75 $\pm$ 0.06 m) participated in the study. Traceurs and basketball players had at least 2 years of training in their particular sports, whereas non-athletes did not have any previous sports background. All participants had no injury in their lower extremity at the time of testing. Following a detailed explanation of the purpose and methodology of the experiments, the subjects gave their written informed consent.

2.2 Protocol

First, passive tissue properties of left AT, TA and GM were measured with a hand-held myotonometer. The participants were asked to lay on a stretcher and relax their body completely for 5 minutes before measurements. After relaxation period, they were asked to lay in a relaxed supine position for measuring TA and, then in a prone position for measuring AT and GM, with their feet hanging off at an ankle angle of 90°.

After myotonometer measurements, the participants were asked to perform 2 types of landing techniques: (1) *Forefoot landing (FFL)*. Only the forefoot touches the ground in plantar flexed position without any subsequent heel contact. (2) *Toe-heel landing (THL)*. The forefoot contact is followed by dorsiflexion until heel contact (Figure 2.1). Note that FFL is a typical technique used by traceurs during landing tasks, whereas THL is widely used by athletes in other fields and non-athletes. In both landing techniques participants instructed to keep their legs parallel to each other in sagittal plane.

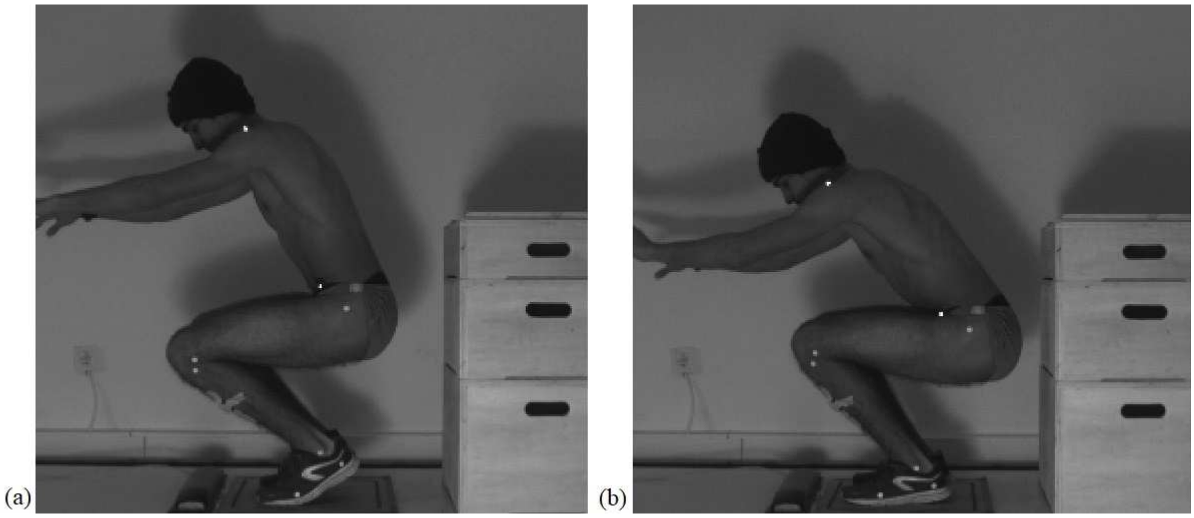

Figure 2.1 FFL (a) and THL (b) techniques.

The participants in different groups were asked to wear the same kind of running shoes at the time of testing. They underwent a warm up protocol including 5 minutes cycling, active stretching and low intensity plyometric work.

The participants were asked to drop off from a 0.75 m high box to land on a force platform (FP) on both feet. To standardize the drop off and to restrain the motion to the vertical direction, the following was done: the participants (i) kept their dominant leg on the box, (ii) moved their non-dominant leg forward and shift their center of mass, (iii) initiate the drop off by a slight push off via their dominant leg and without jumping or bending their knee and, (iv) hold their arms at their sides before, raise them without bending the elbows while, and kept them perpendicular to the ground at the time of landing. After warm up, the participants performed three practice landings for each landing technique for familiarization. During testing, they performed three trials for each landing technique in a randomized order and rested 30 seconds between trials.

2.3 Data Collection

Tissue characteristics of left AT, TA and GM were recorded with a hand-held myotonometer (MyotonPRO, Myoton AS, Tallinn, Estonia). TA and GM measured from the marked sites at muscle belly according to the SENIAM guidelines, and AT measured from a site that is horizontally aligned with the distal portion of lateral malleolus. The probe of MyotonPRO was placed perpendicular to the marked site and the device applied 5 consecutive low-force mechanical impulses (duration: 15 ms; force: 0.58 N) with 1 second intervals. The same procedure was carried out from the same site in each participant for all three measurement sites.

Ground reaction forces (GRF) were collected at 400 Hz from the FP (MatScan, TekScan Inc, South Boston, USA). Piezo-resistive pressure sensors on FP enabled monitoring of the plantar pressure distribution. This system was chosen on purpose to detect heel contact precisely. If heel contact was detected during a FFL trial, that trial was discarded and repeated.

Landing motion was recorded with a high-speed camera at 145 Hz (VDS Vosskühler, CCD-1300QHS, Germany) from left sagittal view. Reflective markers were placed on lateral anatomical reference points at left acromion process, ASIS, greater trochanter, femoral condyle, head of fibula, lateral malleolus, lateral side of the shoe at fifth metatarsal head and heel.

The activity of left TA and GM were collected because of their function in ankle dorsi- and plantar-flexion, which is related to main distinction between the foot placement techniques (THL and FFL) analyzed in this study. The EMG data was collected with a wireless sEMG system (Delsys Trigno Wireless EMG, Delsys, Boston, MA, USA). After the skin was shaved and wiped with alcohol, Delsys Trigno Wireless electrodes with four-contact bars (Material: Ag, Electrode Size: 5 x 1 mm, Inter-electrode Distance: 10 mm) were placed over each muscle according to SENIAM guidelines using double sided tape [84].

The synchronization between three systems was done by a trigger (Trigger module, Delsys, Boston, MA, USA).

2.4 Data Analysis

2.4.1 Ground Reaction Forces

Ground reaction forces were analyzed to obtain information about shock attenuation capacity of each landing technique and group. Using the GRF data analyzed with the software (F-Scan 6.80, TekScan Inc, South Boston, USA), the following metrics were calculated:

(1) *Peak vertical force (pVF)* recorded during landing. pVF was normalized to body weight (BW).

(2) *Time to pVF* from initial contact.

(3) *Loading rate (LR)* i.e., the speed of vertical force increase, was calculated as the ratio of pVF to time to pVF.

Note that, a decreased pVF and LR, and an increased time to pVF indicate enhanced shock attenuation capacity and, thus, a safer landing performance [85].

2.4.2 Joint Kinematics

Anatomical markers were tracked from initial contact (IC) to maximum flexion in the ankle, knee and hip joints. Marker coordinates were filtered through a fourth order Butterworth, low-pass filter with a cutoff frequency of 10 Hz in Matlab (The MathWorks, Inc, Natick, MA, USA). Filtered coordinate data were re-sampled from 145 Hz to 400 Hz, using cubic interpolation. Interpolating the coordinate data to higher sampling rate may enable to better capture maximum joint angles during fast motion [86].

Joint flexion-extension angles and velocities of lower extremities were analyzed to examine the kinematic strategies for shock attenuation. These strategies may provide crucial information about how the joint kinematics affect the shock attenuation process and thus injury risk for each landing technique and group. Using the motion analysis data, ankle, knee and hip joint kinematics were calculated using re-sampled marker coordinates. Each body segment was taken as rigid body and the angles between consecutive body segments (joint angle) were calculated for every frame (Figure 2.2, 2.3, 2.4). The following metrics were calculated for each joint:

(1) *Joint angle at IC* represents joint positions adapted to counter initial impact forces, altering shock attenuation capacity [87, 88],

(2) *joint angle at pVF* represents joint position at the time point of pVF, determining the magnitude and direction of peak GRF acting on the musculoskeletal structures [89, 90],

(3) *joint range from pVF to max flexion* represents total flexion after the time point of pVF, providing information about joint angle regulation in response to pVF [91],

(4) *ROM* represents total flexion throughout landing, altering GRF [23],

(5) *peak angular velocity* represents maximum flexion velocity, providing information about probability of injury occurrence [37, 92],

(6) *mean angular velocity* represents average flexion velocity from IC to max flexion.

(7) In addition, total (dorsi- and plantar-flexion) *ankle angle deviation* was calculated to assess the fast recoil in the ankle angle following IC. This metric may give information about ankle angle control and possible elastic recoil of AT after IC. First, time point of balance for the ankle joint was determined by calculating the first 0.01 sec interval in which mean ankle angular velocity was below 8 deg/sec. *Ankle angle deviation* was determined from IC to time point of balance as the total angular displacement in the ankle, for both dorsi- and plantar-flexion.

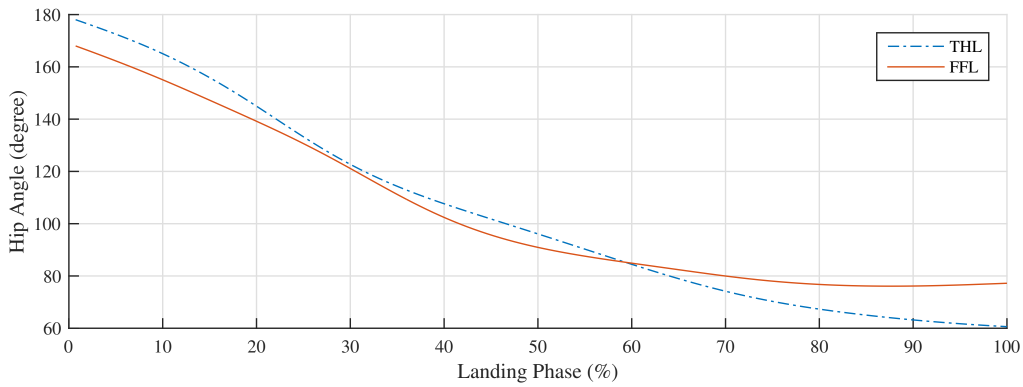

Figure 2.2 Histories of hip angle during a THL and FFL trial.

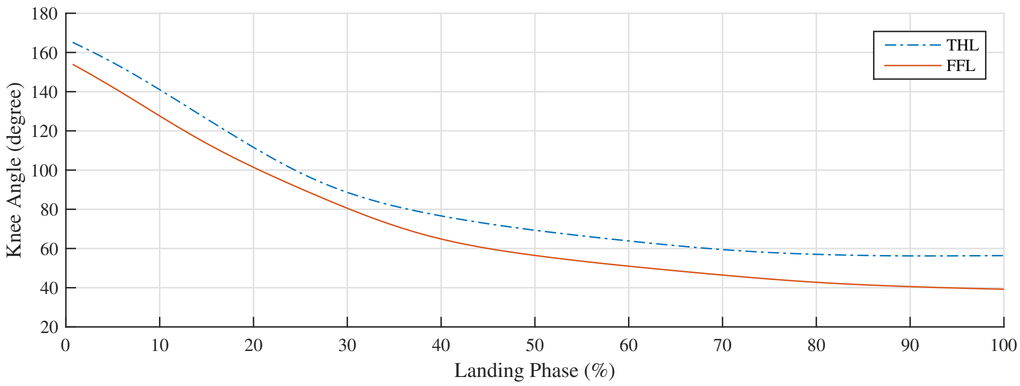

Figure 2.3 Histories of knee angle during a THL and FFL trial.

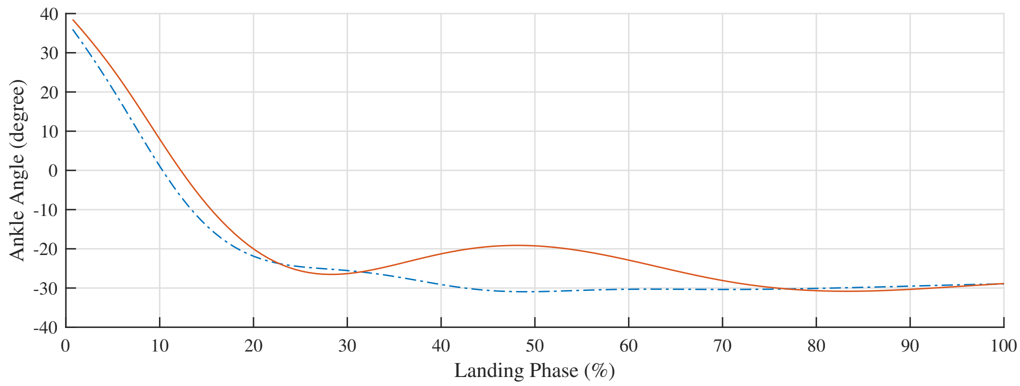

Figure 2.4 Histories of ankle angle during a THL and FFL trial.

2.4.3 Electromyography

The EMG data sampled at 2000 Hz with a bandwidth of 20-450 Hz (Common mode of rejection ratio $>$ 80 dB, Overall Channel Noise $<$ 0.75uV, Resolution: 168 nV / bit) were collected by Trigno EMG system. The EMG signals were rectified and high-pass filtered at 30 Hz (zerolag second-order butterworth) to eliminate motion artifacts caused by the impact of landing [93]. Using rectified and filtered EMG signals, duration of muscle activation from onset to IC (*Activation onset*) was calculated using onset detection method suggested by Santello and McDonagh [33]. This method was chosen because of its reliability in distinguishing between short EMG bursts and continuous EMG activity (Figure 2.5). *Activation onset* defines the point in time that the muscle activation was started and built up continuously in preparation to IC. It provides information about modulation of neuromuscular preparatory mechanisms by CNS for shock absorption [35].

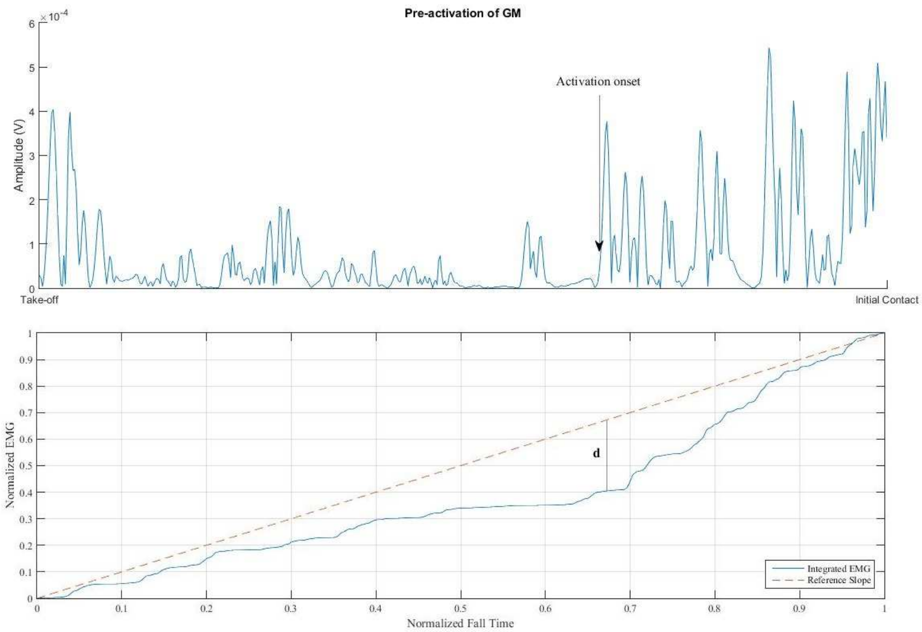

Figure 2.5 Graphical representation of GM pre-activation and method for calculating activation onset. Upper graph shows filtered and fully rectified EMG signal of GM between take-off and IC. Continuous integration was performed for all the data points between take-off and IC (time interval: 2000 Hz). Both time and integrated EMG normalized to 1 in the lower graph. The time point of maximum distance (d) between the integrated EMG and linear reference line (slope = 1), which requires a continuous increase in the EMG amplitude, gives the activation onset.

TA and GM activations were analyzed in 4 phases to obtain information about successive neuromuscular mechanisms during landing: *pre-activation* (100 ms before IC), *post 0-35 activation* (0 to 35 ms after IC), *post 35-80 activation* (35 to 80 ms after IC) and *post 80-200 activation* (80 to 200 ms after IC) (Figure 2.6). The time points and durations of phases were chosen to analyze:

(1) the magnitude of EMG build-up during *pre-activation* (which was reported to start around 120 ms [33]) in preparation to initial impact [33],

(2) the magnitude of continuing yet decreasing pre-programmed muscle activation of the GM and early activation of the TA during *post 0-35* to examine early neuromuscular responses [35],

(3) the magnitude of SLRA of GM and accompanying TA activation between 35 and 80 ms after IC (*post 35-80*) in which peak GRF and musculoskeletal loads are expected to occur [35, 50, 92],

(4) LLRA and voluntary activation of GM, and accompanying TA activation during *post 80-200* to examine late contribution of muscles [39, 40].

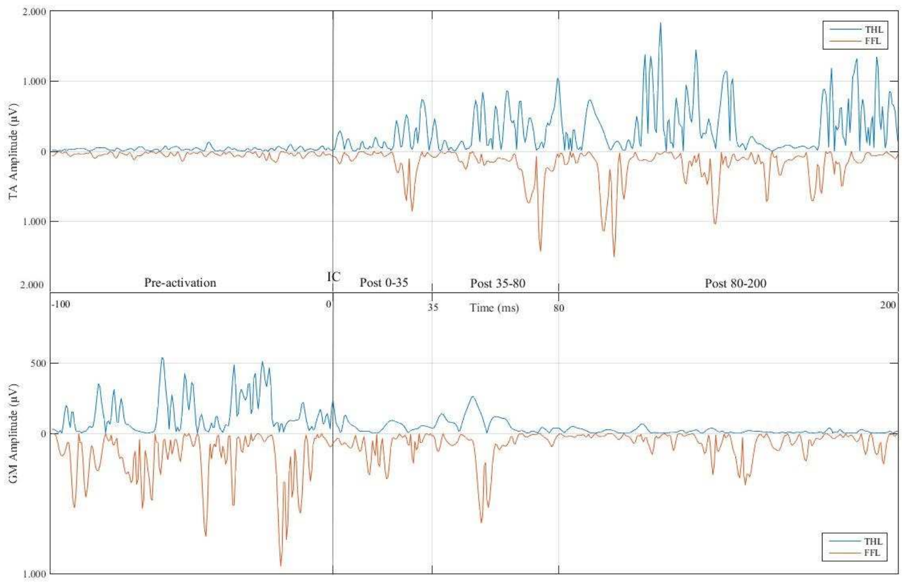

Figure 2.6 Representation of TA and GM pre-, post 0-35, post-35-80 and post 80-200 activation during THL and FFL.

The Root Mean Square (RMS) was calculated for each phase separately according to the Standards for Reporting EMG Data [94]. The EMG signals were not normalized, because (a) normalizing with maximum voluntary isometric contractions (MVIC) represent fast movements poorly [95] and (b) normalizing by within-trial peak or mean amplitudes gives no information about the neuromuscular requirement of the task which is required for comparison of techniques in this study [96]. Therefore, absolute EMG amplitudes were analyzed and used for technique and phase comparisons, inter-subject comparisons were not made. *Co-activation index* between TA and GM was calculated for each phase (Eq. 2.1) [97]. *Co-activation index* may provide important information about the regulation of joint stiffness [41]. EMG amplitudes were normalized for the analysis of *co-activation index* with peak dynamic normalization, a reliable method for the analysis of fast movements, to compare two muscles (TA &

MG) and groups [98]. Additionally, time of maximum amplitudes were calculated for: (i) 0-50 ms (*Post 0-50 max time*) to detect early peak of TA [26], (ii) 0-80 ms post-landing (*Post 0-80 max time*) to detect time point of peak SLRA [26, 32], (iii) 0-200 ms post-landing (*Post 0-200 max time*) to detect time point of peak LLR [26, 40].

$$\textit{Co-activation Index} = 2 \times \frac{\textit{Norm EMG GM}}{\textit{Norm EMG TA} + \textit{Norm EMG GM}} \times 100 \tag{2.1}$$

For each metric, mean values of three trials of each technique were calculated.

2.4.4 Tissue Mechanical Characteristics

Mechanical characteristics of the TA and GM muscles, and the AT were analyzed to examine the mechanical adaptations in lower leg muscles and tendon in response to habitual sport specific training and foot placement technique. Accelerometers within MyotonPRO sampled (3200 Hz) the damped natural oscillations caused by the probe impact (Figure 2.7).

The following metrics were calculated:

(1) *frequency (Hz)* represents oscillation frequency of the tissue and it characterizes the muscle tone. An increase in the measured frequency indicates an elevated muscle tone (Eq. 2.2).

$$F = \mathrm{f\ \_max} \tag{2.2}$$

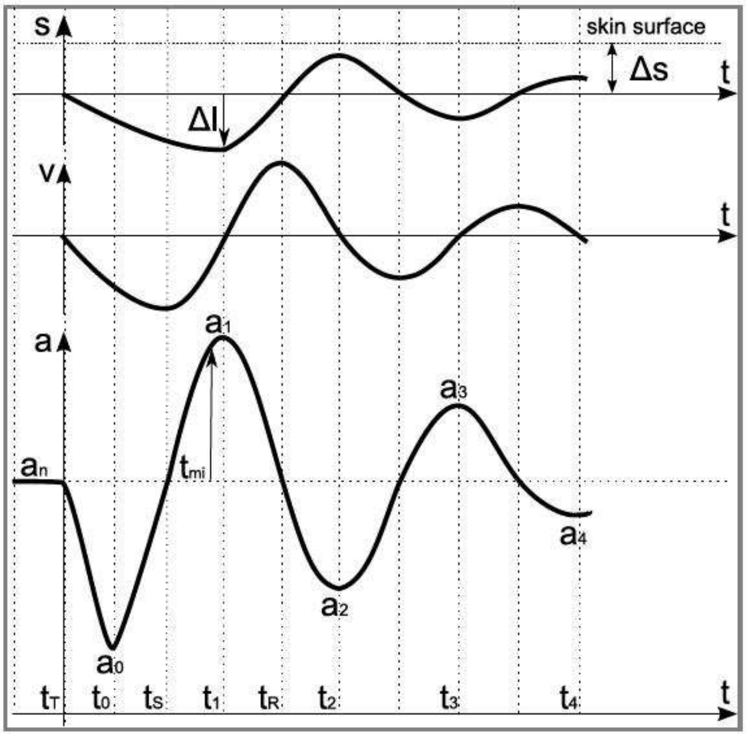

Figure 2.7 Representation of displacement-, oscillation velocity-, acceleration-time graphs of natural oscillations of tissue after MyotonPro impact. Δl: maximum displacement of the tissue, t_T: Start of the mechanical impulse, a_1: Maximum displacement in maximum tissue resistance measured in mG, t_1: The time when maximum displacement was reached, t_r: The time when tissue returns to its initial shape, a_3: Maximum displacement of the second period of oscillation which takes place due to the recuperation of stored residual mechanical energy in the tissue [99].

(2) *stiffness (N/m)* represents dynamic stiffness of the tissue. An increase in the measured stiffness indicates an elevated resistance to deformation (Eq. 2.3).

$$S = \frac{\mathrm{a\ \_max} \times \mathrm{m\ \_probe}}{\Delta l} \tag{2.3}$$

(3) *elasticity (logarithmic decrement)* represents the ability of the tissue to recover its initial shape. An increased logarithmic decrement measured indicates a lower elasticity and a higher mechanical energy loss during deformation (Eq. 2.4).

$$D = ln\left(\frac{a_1}{a_3}\right) \tag{2.4}$$

(4) *relaxation (ms)* represents the mechanical stress relaxation time of the tissue after an external (such as impact) or internal (such as contraction) deformation. An increased relaxation time measured indicates that the tissue fully recovers its shape only slowly after deformation (Eq. 2.5).

$$R = t_R - t_1 \tag{2.5}$$

(5) *creep* represents the ratio of relaxation time to deformation time of the tissue. An increased creep rate indicates that the time difference between relaxation and deformation is relatively small (Eq. 2.6) [99].

$$C = \frac{R}{t_1 - t_t} \tag{2.6}$$

The mean value of 5 consecutive measurements were calculated and used for analyses.

2.5 Statistical Analysis

Two-way ANOVA for repeated measures (factors: group and technique) was performed to analyze the effects of sports training and technique on joint kinematics and GRF metrics of landing. Paired samples T-test was used to compare EMG activity onset and max times between techniques in each group. Multivariable ANOVA (factors: phase and technique) was used to assess the within-subject differences in the EMG amplitudes between two techniques and phases, as well as interactions of the technique with each phase. One-way ANOVA was used to compare the differences in mechanical characteristics of the AT, TA and GM between groups. If significant main effects were found, Tukey (main effects) and Bonferroni (interactions) post-hoc tests were conducted to further locate significant within factor differences. If no significant interactions were found, average changes were reported for each metric. The level of significance was set at $P < 0.05$.

3. RESULTS

Reliability analyses were performed to determine if there were any significant differences between trials because of fatigue or any other factor. Cronbach's alpha was calculated to comparing the 3 trials of both landing techniques for each studied parameter. The Cronbach's alpha was over 0.70 in 39 of the 44 GRF and kinematic variables (2 techniques x 22 metrics) and over 0.60 in 5 variables. Therefore, reliability analyses presented no significant difference between trials, so mean values of 3 trials were calculated for further analyses for each landing metric.

3.1 Ground Reaction Forces

Mean (SD) values of each GRF metric are shown in Table 3.1 for all groups and techniques.

Table 3.1
Group means (SD) of ground reaction force metrics for each technique.

	Basketball		**Traceur**		**Non-athlete**	
	THL	**FFL**	**THL**	**FFL**	**THL**	**FFL**
$pVF\ (BW)^{g\ \&\ t}$	4.85 (0.57)	3.38 (0.35)	4.27 (0.59)	3.04 (0.30)	4.68 (0.74)	3.08 (0.38)
$Time\ to\ pVF\ (sec)^{t}$	0.043 (0.005)	0.068 (0.013)	0.044 (0.005)	0.069 (0.012)	0.040 (0.006)	0.069 (0.013)
$LR\ (BW/sec)^{t\ \&\ i}$	114.64 (19.18)	52.22 (13.67)	98.94 (13.25)	46.40 (13.05)	121.12 (25.39)	46.85 (12.40)

^{g} Significant main effects of group, ^{t} Significant main effects of technique, ^{i} Significant interaction

ANOVA (factors: group and technique) showed:

pVF significant main effects of both factors ($P<0.05$), and no significant interaction. pVF was higher (by 1.43 BW, 45.2%) for THL. *Post-hoc*: no significant difference between groups for each technique ($P>0.05$).

time to pVF significant main effects of only technique ($P<0.001$), and no significant interaction. The mean time to pVF was higher (by 0.27 sec, 39.1%) for FFL which indicates a slower increase in vertical ground reaction forces from ground contact to pVF.

LR significant main effects of only technique ($P<0.001$), but a significant interaction ($P<0.05$). The mean LR was higher by 62.41 BW/sec (119.5%) in basketball players, 52.54 BW/sec (113.2%) in traceurs and 74.28 BW/sec (158.5%) in non-athletes for THL.

FFL technique demonstrated significantly lower pVF, LR and slower time to pVF values, indicating a safer and more efficient shock absorption with this foot placement technique [85, 89, 100, 101]. There was no significant difference found between groups for each technique and metric, including pVF, for which only significant main effects of group was found but no significance in within technique comparisons. Significant interaction of technique and group was found for LR; non-athletes demonstrated higher LR than traceurs during THL, however the difference disappeared when performing FFL, indicating FFL was similarly effective for shock attenuation in non-athletes.

3.2 Joint Kinematics

Mean (SD) values of each kinematic metric for hip, knee, and ankle joints are shown in Table 3.2 and Table 3.3 for all groups and techniques.

Table 3.2
Group means (SD) of joint angle at IC, joint angle at pVF and joint range from pVF to max flexion for each technique.

	Basketball		**Traceur**		**Non-athlete**	
	THL	**FFL**	**THL**	**FFL**	**THL**	**FFL**
Joint angle at IC (°)						
Hip ^{NS}	159.04 (6.34)	159.02 (6.68)	160.79 (7.27)	157.76 (6.71)	159.64 (7.67)	160.29 (6.93)
Knee ^{t}	154.57 (3.35)	151.07 (4.03)	152.95 (6.86)	147.34 (5.96)	155.37 (6.77)	150.78 (6.51)
Ankle ^{t}	27.76 (4.39)	36.44 (4.21)	27.57 (6.83)	32.55 (6.38)	29.34 (2.90)	36.62 (3.64)
Joint angle at pVF (°)						
Hip ^{t}	146.24 (5.57)	137.65 (5.59)	145.14 (10.05)	133.33 (10.53)	147.44 (6.37)	137.66 (9.12)
Knee ^{t}	129.85 (3.78)	113.07 (5.98)	124.82 (9.02)	106.08 (10.08)	130.92 (6.21)	110.13 (10.86)
Ankle ^{t}	-8.00 (3.74)	-11.96 (4.40)	-12.04 (6.32)	-14.27 (9.82)	-7.17 (5.33)	-13.57 (7.30)
Joint range from pVF to max flexion (°)						
Hip ^{t}	74.70 (16.64)	61.95 (6.87)	90.64 (17.33)	71.04 (18.07)	91.60 (18.07)	69.88 (16.03)
Knee $^{g\ \&\ t}$	55.51 (6.82)	46.77 (6.66)	63.74 (13.75)	50.54 (9.93)	76.66 (16.76)	60.04 (14.37)
Ankle $^{t\ \&\ i}$	16.84 (2.30)	4.24 (2.17)	18.26 (5.26)	4.25 (3.28)	22.85 (6.92)	3.27 (4.40)

For the ankle angle, neutral position = 0°and negative values represent dorsi-flexion. ^{NS} No significant main effect and interaction, ^{g} Significant main effects of group, ^{t} Significant main effects of technique, ^{i} Significant interaction

Table 3.3
Group means (SD) of joint ROM, peak velocity, mean velocity and ankle angle deviation for each technique.

	Basketball		**Traceur**		**Non-athlete**	
	THL	**FFL**	**THL**	**FFL**	**THL**	**FFL**
ROM (°)						
Hip $^{g\ \&\ t}$	87.49 (14.67)	83.32 (7.62)	106.29 (15.46)	95.47 (15.54)	103.80 (17.92)	92.52 (14.06)
Knee ^{g}	80.22 (6.35)	84.77 (7.99)	91.88 (11.56)	91.80 (9.46)	101.12 (15.87)	100.69 (9.81)
Ankle $^{t\ \&\ i}$	52.60 (4.72)	52.64 (4.20)	57.87 (4.78)	51.06 (6.39)	59.36 (7.48)	53.46 (4.80)
Peak Velocity (°/sec)						
Hip $^{g\ \&\ t}$	574.78 (62.18)	511.63 (77.27)	629.56 (31.42)	557.02 (53.43)	642.35 (73.00)	567.49 (77.08)
Knee ^{t}	718.95 (56.29)	631.54 (81.33)	744.27 (60.17)	660.46 (82.52)	797.21 (70.63)	669.25 (82.65)
Ankle ^{t}	968.46 (96.02)	954.83 (115.20)	1017.31 (108.97)	929.66 (109.91)	1026.39 (76.15)	1000.84 (65.80)
Mean Velocity (°/sec)						
Hip $^{g\ \&\ t}$	219.59 (31.36)	208.58 (35.23)	268.73 (32.83)	251.71 (35.61)	245.22 (33.65)	225.25 (24.57)
Knee ^{t}	250.47 (36.61)	236.65 (42.85)	284.83 (50.36)	266.15 (44.45)	277.31 (63.03)	250.68 (44.60)
Ankle $^{t\ \&\ i}$	227.36 (52.36)	588.76 (73.30)	265.33 (55.69)	561.08 (93.69)	196.49 (58.18)	628.41 (38.15)
Ankle Angle Deviation (°) $^{g\ \&\ t}$	51.62 (4.01)	63.86 (5.80)	57.51 (5.27)	62.34 (7.24)	58.67 (9.01)	68.63 (7.83)

For the ankle angle, neutral position= 0°and negative values represent dorsi-flexion. ^{NS} No significant main effect and interaction, ^{g} Significant main effects of group, ^{t} Significant main effects of technique, ^{i} Significant interaction

ANOVA (factors: group and technique) showed:

Joint angle at IC

Hip: no significant main effects and interaction. **Knee:** significant main effects of only technique (P<0.001), and no significant interaction. The knee was more extended (by 4.57°, 3.1%) for THL. **Ankle:** significant main effects of only technique (P<0.001), and no significant interaction. The ankle was more dorsi-flexed (by 6.98°, 19.8%) for THL.

Joint angle at pVF

Hip: significant main effects of only technique (P<0.001), and no significant interaction. The hip was more extended (by 10.06°, 7.4%) for THL. **Knee:** significant main effects of only technique (P<0.001), and no significant interaction. The knee was more extended (by 18.77°, 17.1%) for THL. **Ankle:** significant main effects of only technique (P<0.001), and no significant interaction. The ankle was more plantar-flexed (by 4.19°, 31.6%) for THL.

Joint range from pVF to max flexion

Hip: significant main effects of only technique (P<0.001), and no significant interaction. The hip was more flexed (by 18.04°, 26.6%) for THL. **Knee:** significant main effects of technique (P<0.001) and group (P<0.05), but no significant interaction. The knee was more flexed (by 12.85°, 24.5%) for THL. *Post-hoc*: basketball players flexed their knee less than non-athletes (by 21.16°, 27.6% and 13.27°, 22.1% for THL and FFL, respectively). **Ankle:** significant main effects of only technique (P<0.001), but a significant interaction (P<0.05). The ankle was dorsi-flexed more by 12.6°(up to 3-fold) in basketball players, 14.0°(up to 3-fold) in traceurs and 19.60 (up to 6-fold) in non-athletes for THL.

ROM

Hip: significant main effects of technique ($P<0.001$) and group ($P<0.05$), but no significant interaction. The hip was flexed more (by 15.40°, 9.7%) for THL. *Post-hoc*: basketball players flexed their hip less than traceurs (by 18.80°, 17.7% and 12.15°, 12.7% for THL and FFL, respectively). **Knee:** significant main effects of only group ($P<0.001$), and no significant interaction. *Post-hoc*: basketball players flexed their knee less than non-athletes (by 20.89°, 20.7% and 15.92°, 15.8% for THL and FFL, respectively). **Ankle:** significant main effects of only technique ($P<0.001$), but a significant interaction. The ankle was dorsi-flexed less by 0.037°(0.1%) in basketball players, whereas it was dorsi-flexed more by 6.81°(13.3%) in traceurs and 5.900 (11.0%) in non-athletes for THL.

Peak Velocity

Hip: significant main effects of technique ($P<0.001$) and group ($P<0.05$), but no significant interaction. Hip peak velocity was higher (by 70.18°/sec, 12.9%) for THL. *Post-hoc*: basketball players' hip peak velocity was lower than non-athletes (by 67.57°/sec, 10.5% and 55.86°/sec, 9.8% for THL and FFL, respectively). **Knee:** significant main effects of only technique ($P < 0.001$), and no significant interaction. The knee peak velocity was higher (by 99.73°/sec, 15.3%) for THL. **Ankle:** significant main effects of only technique ($P < 0.05$), and no significant interaction. The ankle peak velocity was higher (by 42.28°/sec, 4.4%) for THL.

Mean Velocity

Hip: significant main effects of technique ($P<0.05$) and group ($P<0.05$), but no significant interaction. Hip mean velocity was higher (by 16.00°/sec, 7.0%) for THL. *Post-hoc*: basketball players' hip mean velocity was lower than that of traceurs (by 49.14°/sec, 18.3% and 43.13°/sec, 17.1% for THL and FFL, respectively). **Knee:** significant main effects of only technique ($P<0.05$), and no significant interaction. The knee mean velocity was higher (by 19.71°/sec, 7.8%) for THL. **Ankle:** significant main effects of only technique (P<0.01), but a significant interaction (P<0.05). The ankle mean velocity was lower by 361.40°/sec (61.4%) in basketball players, 295.76°/sec (52.7%) in traceurs and 431.920/sec (68.7%) in non-athletes for THL.

Ankle Angle Deviation

ANOVA (factors: group and technique) showed significant main effects of group (P<0.05) and technique (P<0.001), but no significant interaction. Ankle angle deviation was lower (by 9.000, 13.9%) for THL. *Post-hoc*: basketball players' ankle angle deviation was lower than non-athletes (by 7.05°, 12.0% and 4.77°, 6.9% for THL and FFL, respectively).

Therefore, overall, the kinematic findings revealed the following: (a) FFL led to pre-programmed joint positions at IC, resulting in a more efficient and safer shock attenuation, which agrees with previous reports [87, 88, 102]. (b) The later occurrence of pVF during FFL have led to more flexed, safer joint positions at pVF [90, 92, 103]. (c) Higher ROM and velocity values measured during THL indicate a compensation mechanism for the increased GRF [37, 91]. (d) Basketball players flexed their joints less than other groups, which may indicate a specific adaptation to sports' requirements such as higher leg stiffness and restricted dorsiflexion ROM [49, 78, 104–107]. (e) The traceurs exhibited the highest mean dorsiflexion velocity during THL and a lower dorsiflexion velocity for FFL, which may indicate increased efficiency in control of the ankle during landing ascribed to their habitual foot placement technique (i.e., FFL). (f) The higher ankle angle deviation during FFL may indicate utilization of spring like properties of the AT to a greater extent than in THL, increasing energy dissipation in the ankle joint [108, 109].

3.3 Electromyography

Mean (SD) values of activation onsets and max times for left TA and GM muscles are shown in Table 3.4 for all groups, techniques and phases. Mean values were not calculated and showed on the table for amplitudes because only within subject differences and interactions were analyzed for these. Since no normalization was done, averaging would be methodologically wrong.

Table 3.4
Group means (SD) of EMG metrics for each technique.

	Basketball		**Traceur**		**Non-athlete**	
	THL	**FFL**	**THL**	**FFL**	**THL**	**FFL**
Activation Onset (ms)						
TA ^{NS}	98.20 (21.70)	106.42 (20.50)	99.82 (20.84)	104.04 (17.08)	92.31 (20.34)	106.30 (12.37)
GM ^{NS}	115.94 (13.60)	118.17 (13.04)	117.51 (19.30)	120.62 (18.54)	124.03 (23.42)	122.15 (17.70)
Post 0-50 Max Time (ms)						
TA ^{NS}	27.89 (8.49)	28.83 (8.59)	28.35 (7.05)	30.79 (4.88)	30.29 (11.25)	33.92 (8.01)
Post 0-80 Max Time (ms)						
TA ^{NS}	53.17 (13.42)	54.62 (20.99)	58.74 (14.58)	48.06 (14.20)	55.00 (18.58)	49.56 (16.43)
GM ^{NS}	49.35 (4.09)	53.36 (6.05)	57.25 (7.40)	49.94 (7.26)	53.53 (6.36)	51.11 (9.64)
Post 0-200 Max Time (ms)						
TA ^{t}	135.18 (23.31)	113.92 (28.82)	118.33 (44.91)	89.22 (38.18)	113.60 (43.30)	92.29 (38.87)
GM ^{t}	57.01 (21.41)	105.00 (40.65)	74.29 (25.86)	102.78 (47.86)	59.55 (21.91)	114.96 (38.29) 100.69)

For the activation onset and pre max times, values are given as the duration before IC. For the post max times, values are given as the duration after IC. ^{NS} No significant difference between techniques, ^{t} Significant difference between techniques

Paired samples T-test showed following for EMG activation onset and max times:

Activation Onset

TA: no significant difference between techniques ($P>0.05$). **GM:** no significant difference between techniques ($P>0.05$).

Pre Max Time

TA: significant difference of technique ($P<0.05$) only in non-athletes. TA activation peaked earlier (by 22.1 ms, 56.2%) for FFL in non-athletes. **GM:** no significant difference between techniques ($P>0.05$).

Post 0-50 Max Time

TA: no significant difference between techniques ($P>0.05$).

Post 0-80 Max Time

TA: no significant difference between techniques ($P>0.05$). **GM:** no significant difference between techniques ($P>0.05$).

Post 0-200 Max Time

TA: significant difference of technique ($P<0.05$) in all groups. TA activation peaked later by 21.3 ms (18.4%) in basketball players, 29.1 ms (32.5%) in traceurs and 21.3 ms (22.8%) in non-athletes for THL. **GM:** significant difference of technique ($P<0.05$) in basketball players and non-athletes. GM activation peaked later by 48.0 ms (84.2%) in basketball players and 55.4 ms (93.0%) in non-athletes for FFL.

ANOVA (factors: technique and phase) showed the following for the EMG amplitudes:

TA Amplitude

Significant main effects of technique and phase, and a significant interaction in each group ($P<0.05$). The change in TA amplitude throughout the landing period is illustrated for each group (Figure 3.1, 3.2, 3.3). Post-hoc tests revealed:

Basketball players: an elevated activation for the THL ($P<0.05$) during post 0-35 (by 23.1 μV, 20.6%), 35-80 (by 52.8 μV, 42.7%) and 80-200 (by 85.3 μV, 55.3%) compared to FFL. TA activation was significantly increased from pre to post 0-35 and post 35-80 to 80-200 for THL; only from pre to post 35 for FFL ($P<0.05$).

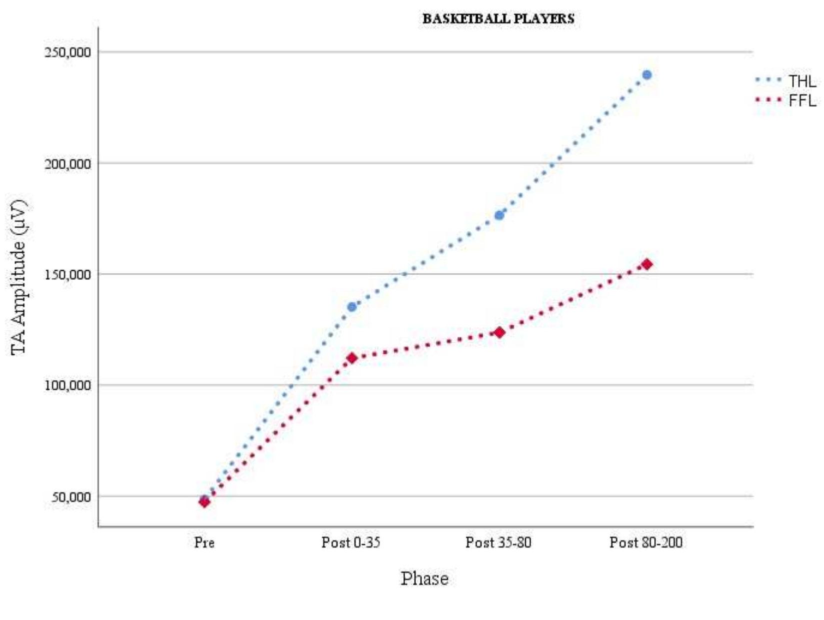

Figure 3.1 Representation of basketball players' mean TA amplitude during THL and FFL.

Traceurs: an elevated activation for the THL ($P<0.05$) during 35-80 (by 40.5 μV, 27.6%) and 80-200 (by 66.8 μV, 44.8%) compared to FFL. TA activation was significantly increased from pre to post 0-35 and post 0-35 to 35-80 for THL; only from pre to post 35 for FFL ($P<0.05$).

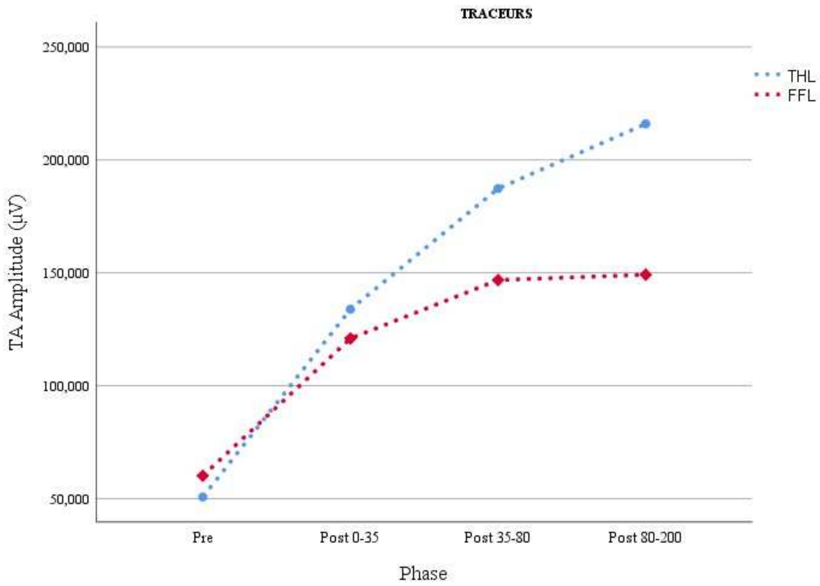

Figure 3.2 Representation of traceurs' mean TA amplitude during THL and FFL.

Non-athletes: an elevated activation for the THL (P<0.05) during 35-80 (by 45.3 μV, 37.1%) and 80-200 (by 59.4 μV, 42.7%) compared to FFL. TA activation was significantly increased only from pre to post 35 for both THL and FFL (P<0.05).

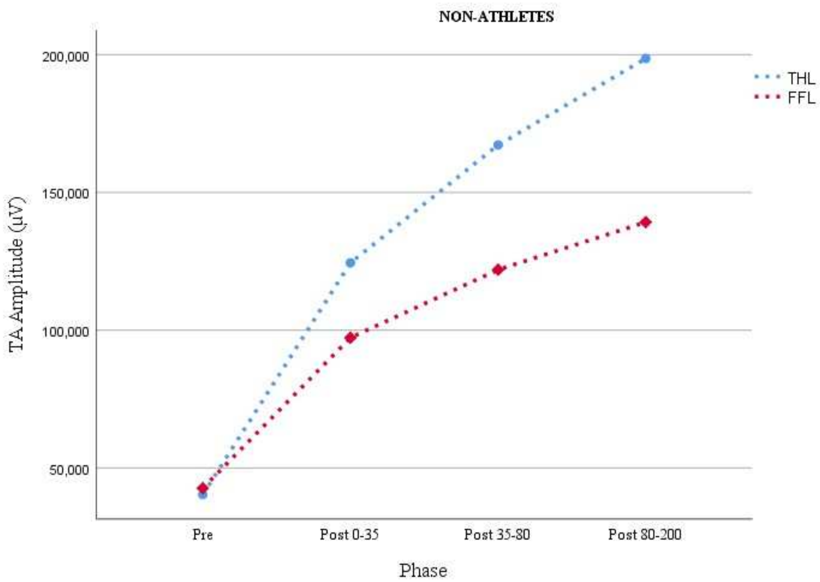

Figure 3.3 Representation of non-athletes' mean TA amplitude during THL and FFL.

GM Amplitude

Significant main effects of technique and phase, and a significant interaction were shown in each group (P<0.05). The change in GM amplitude throughout the landing period is illustrated for each group (Figure 3.4, 3.5, 3.6). Post-hoc tests revealed:

Basketball players: an elevated activation for the FFL (P<0.05) during pre (by 33.7 μV, 24.8%), post 0-35 (by 26.9 μV, 45.3%) and 80-200 (by 27.3 μV, 171.0%) compared to THL. GM activation was significantly decreased from pre to post 0-35 and post 35-80 to 80-200 for both THL and FFL (P<0.05).

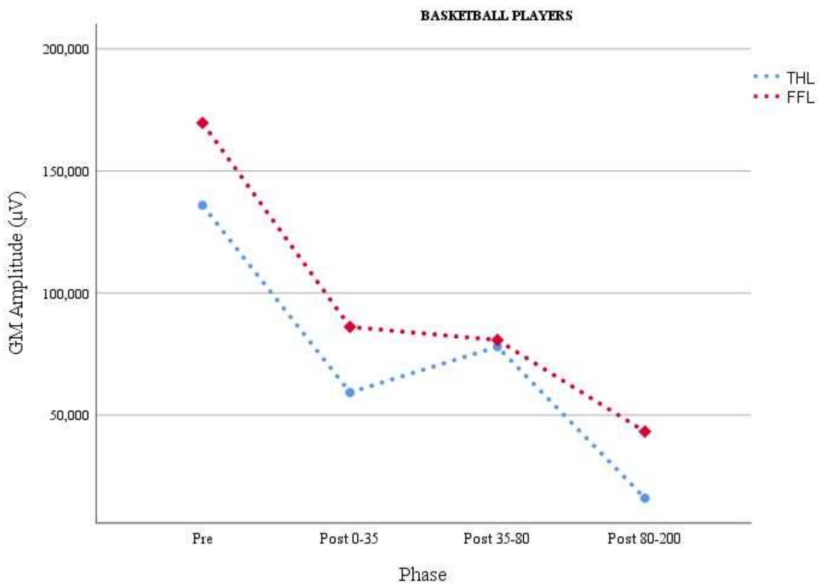

Figure 3.4 Representation of basketball players' mean GM amplitude during THL and FFL.

Traceurs: an elevated activation for the FFL (P<0.05) during pre (by 18.7 μV, 15.5%), post 0-35 (by 33.5 μV, 78.6%), 35-80 (by 18.4 μV, 34.3%) and 80-200 (by 21.5 μV, 109.5%) compared to THL. GM activation was significantly decreased only from post 35-80 to 80-200 for both THL and FFL (P<0.05).

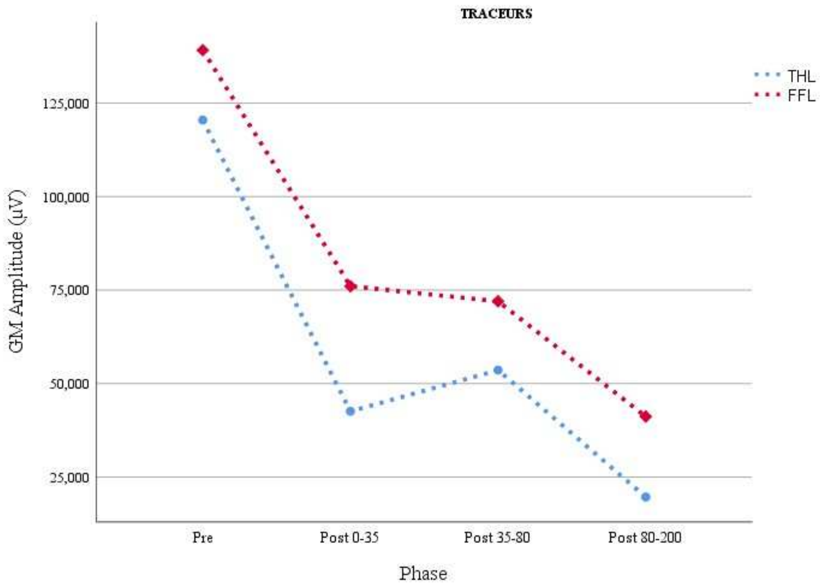

Figure 3.5 Representation of traceurs' mean GM amplitude during THL and FFL.

Non-athletes: an elevated activation for the FFL (P<0.05) during pre (by 27.3 μV, 26.9%), post 0-35 (by 26.8 μV, 75.7%) and 80-200 (by 25.6 μV, 256.6%) compared to THL. GM activation was significantly decreased only from pre to post 0-35 and post 35-80 to 80-200 for THL; and pre to post 35, post 0-35 to 35-80 and post 35-80 to 80-200 for FFL (P<0.05).

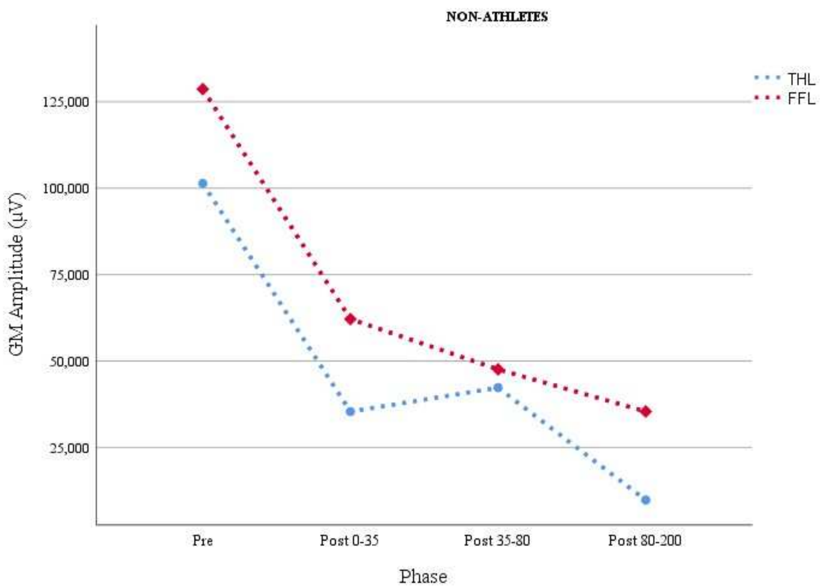

Figure 3.6 Representation of non-athletes' mean GM amplitude during THL and FFL.

Co-activation index:

Significant main effects of phase and a significant interaction were shown in each group, and significant main effects of technique were shown only in non-athletes ($P<0.05$). The change in co-activation throughout landing period illustrated for each group (Figure 3.7, 3.8, 3.9). ANOVA (factors: technique and phase) showed:

Basketball players: No significant main effects of technique ($P>0.05$). However, post-hoc revealed significant differences between techniques within phases: Pre co-activation was significantly higher (by 5.7%) ($P<0.05$), whereas post 80-200 co-activation was significantly lower (by 47.9%) ($P<0.001$) for THL. Co-activation was significantly decreased from pre to post 0-35 and post 35-80 to 80-200 for THL ($P<0.001$) (by 42.9% and 72.9%, respectively) and FFL (by 40.5% and 42.6%, respectively) ($P<0.01$).

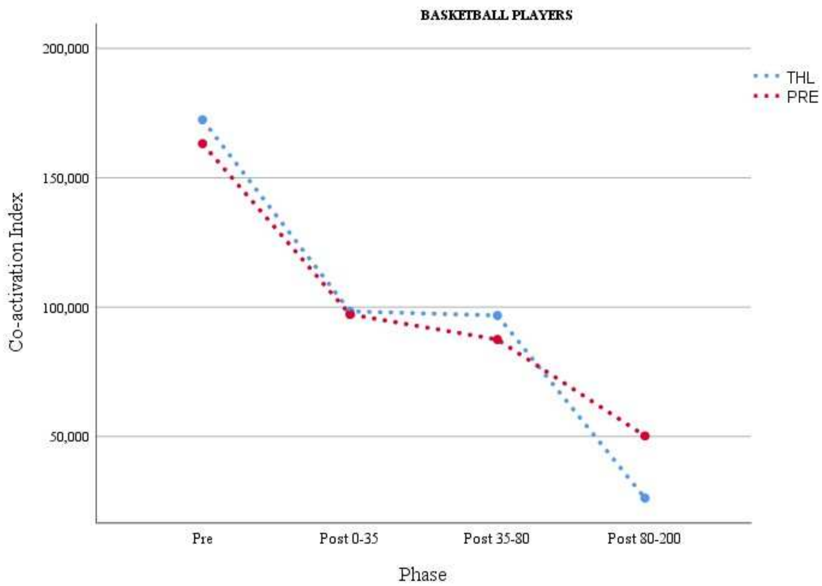

Figure 3.7 Representation of basket players' mean co-contraction index during THL and FFL.

Traceurs: No significant main effects of technique ($P>0.05$). However, post-hoc revealed significant differences between techniques within phases: Pre co-activation was significantly higher (by 12.7%) ($P<0.05$), whereas post 80-200 co-activation was significantly lower (by 42.6%) ($P<0.001$) for THL. Co-activation was significantly decreased from pre to post 0-35 and post 35-80 to 80-200 for THL (by 50.6% and 57.4%, respectively), and only from pre to post 0-35 for FFL (by 37.4%) (P<0.001).

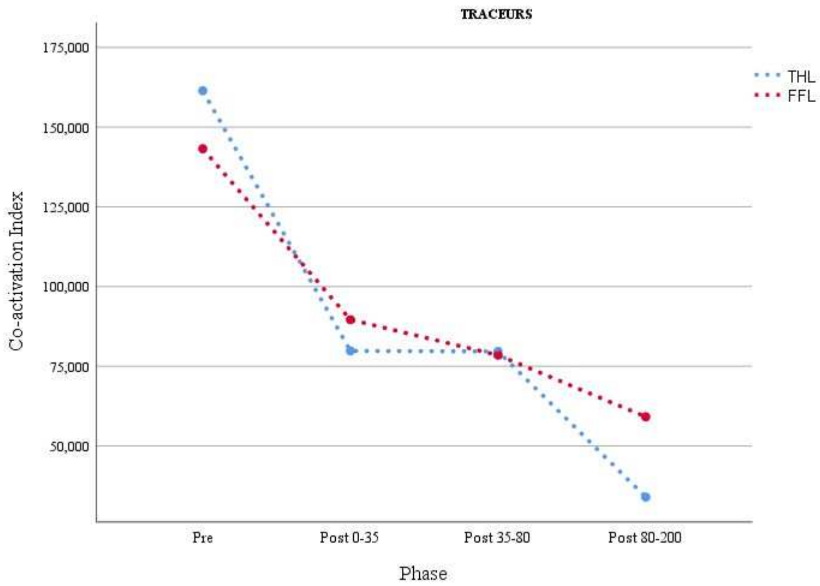

Figure 3.8 Representation of traceurs' mean co-contraction index during THL and FFL.

Non-athletes: significant main effects of technique (P<0.05). Post-hoc revealed significant differences between techniques within phases: Pre co-activation was significantly higher (by 5.8%) (P<0.01), whereas post 80-200 co-activation was significantly lower (by 62.6%) (P<0.001) for FFL. Co-activation was significantly increased from pre to post 0-35 and post 35-80 to 80-200 for THL (by 53.6% and 69.6%, respectively), and only from pre to post 0-35 for FFL (by 35.3%) (P<0.001).

Overall, the EMG findings revealed the following: (a) Muscles' activation onsets did not differ between techniques, indicating an ingrained mechanism for activation onset modulation. (b) Considering that the onsets did not differ between techniques [35], the higher GM pre-activation levels for the FFL indicate a higher rate of GM EMG built-up. (c) Similar SLRA peak times of the GM (∼53 and ∼52 ms after IC for THL and FFL, respectively) between techniques, in the presence of different times to pVF (∼42 and ∼69 ms after IC for THL and FFL, respectively) indicate that the FFL may reduce the injury risk by allowing sufficient time for plantar flexors to respond to pVF [92, 110, 111]. (d) Differences in post-activation trends of the TA and GM for

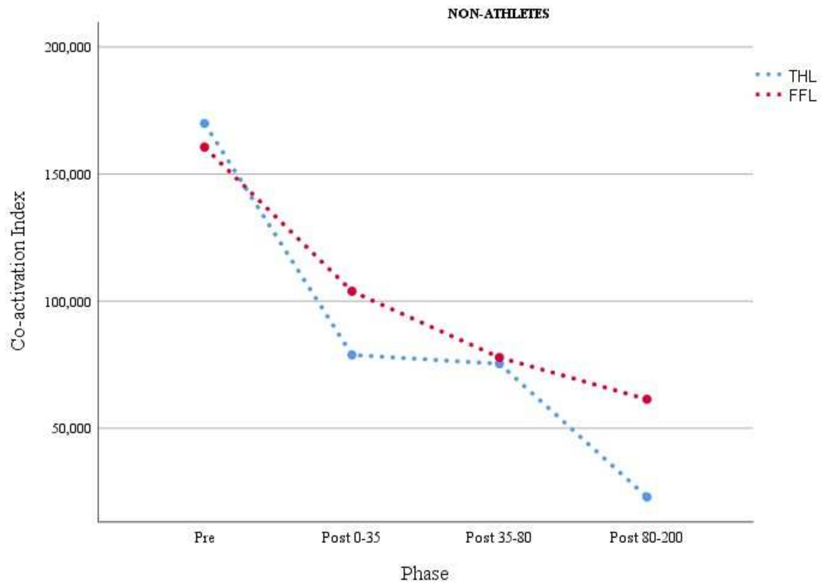

Figure 3.9 Representation of non-athletes' mean co-contraction index during THL and FFL.

each technique may indicate a regulation mechanism to maintain balance, in response to anterioposterior position differences of COM between techniques [76]. (e) Co-activation was the highest before IC to stiffen and stabilize the ankle joint in preparation to initial impact, and it gradually decreased throughout landing [41,112]. (f) Longer and higher activation of plantar flexors during the FFL may keep muscle fascicles close to their optimum lengths and increase their stiffness, enhancing the spring like behavior of the AT during ankle deviation [35,109,113–115].

3.4 Tissue Mechanical Characteristics

Mean (SD) values of each mechanical characteristic for left TA, GM and AT are shown in Table 3.5 and Table 3.6 for all groups, techniques and phases.

Table 3.5
Group means (SD) of frequency and stiffness of TA, GM and AT.

	Basketball	**Traceur**	**Non-athlete**
Frequency (Hz)			
TA ^{g}	22.15 (1.14)	22.12 (1.84)	20.32 (1.96)
GM ^{g}	18.87 (1.54)	19.45 (1.78)	17.64 (1.34)
AT ^{g}	38.52 (4.09)	38.99 (3.68)	35.18 (1.96)
Stiffness (N/m)			
TA ^{g}	470.08 (46.21)	477.00 (54.26)	414.17 (72.06)
GM ^{g}	368.33 (57.19)	396.33 (52.75)	335.92 (38.81)
AT ^{g}	1071.67 (83.98)	1087.08 (63.14)	996.25 (62.97)

^{NS} No significant difference between groups, ^{g} Significant difference between groups

One-way ANOVA (factor: group) showed the following:

TA

Frequency: significant difference between groups (P<0.05). Post-hoc: Basketball players' and traceurs' TA tone was significantly higher than non-athletes by 9.0% and 8.9%, respectively (P<0.05). **Stiffness:** significant difference between groups (P<0.05). Post-hoc: Traceurs' TA stiffness was significantly higher than non-athletes by 15.2% (P<0.05). **Elasticity:** no significant difference between groups (P>0.05). **Relaxation:** significant difference between groups (P<0.05). Post-hoc: Traceurs' TA relaxation time was significantly lower than non-athletes by 15.1% (P<0.05). **Creep:** significant difference between groups (P<0.05). Post-hoc: Traceurs' TA creep was significantly lower than non-athletes by 14.5% (P<0.05).

Table 3.6
Group means (SD) of elasticity, relaxation and creep of TA, GM and AT.

	Basketball	**Traceur**	**Non-athlete**
Elasticity			
TA ^{NS}	0.7533 (0,0793)	0.7133 (0.1076)	0,7100 (0.0903)
GM ^{NS}	0.8450 (0.1010)	0.7900 (0.1664)	0.8125 (0.0921)
AT ^{g}	0.3867 (0.0868)	0.3542 (0.1009)	0.4842 (0.1091)
Relaxation (ms)			
TA ^{g}	11.57 (1.26)	11.20 (1.49)	13.19 (2.42)
GM ^{g}	13.87 (1.75)	12.87 (1.75)	14.92 (1.68)
AT ^{NS}	4.57 (0.54)	4.70 (0.79)	5.05 (0.44)
Creep			
TA ^{g}	0.7375 (0.0789)	0.7100 (0.0889)	0.8300 (0.1422)
GM ^{g}	0.8517 (0.0991)	0.7917 (0.0995)	0.9075 (0.0973)
AT ^{NS}	0.3167 (0.0337)	0.3450 (0.1111)	0.3500 (0.0245)

^{NS} No significant difference between groups, ^{g} Significant difference between groups

GM

Frequency: significant difference between groups (P<0.05). Post-hoc: Traceurs' GM tone was significantly higher than non-athletes by 10.2% (P<0.05). **Stiffness:** significant difference between groups (P<0.05). Post-hoc: Traceurs' GM stiffness was significantly higher than non-athletes by 18.0% (P<0.05). **Elasticity:** no significant difference between groups (P>0.05). **Relaxation:** significant difference between groups (P<0.05). Post-hoc: Traceurs' GM relaxation time was significantly lower than non-athletes by 13.7% (P<0.05). **Creep:** significant difference between groups

(P<0.05). Post-hoc: Traceurs' GM creep was significantly lower than non-athletes by 12.8% (P<0.05).

AT

Frequency: significant difference between groups (P<0.05). Post-hoc: Traceurs' AT tone was significantly higher than non-athletes by 10.8% (P<0.05). **Stiffness:** significant difference between groups (P<0.05). Post-hoc: Basketball players' and traceurs' AT stiffness was significantly higher than non-athletes by 7.6% and 9.1%, respectively (P<0.05). **Elasticity:** significant difference between groups (P<0.05). Post-hoc: Traceurs' AT elasticity was significantly higher than non-athletes by 36.7% (P<0.01). **Relaxation:** no significant difference between groups (P>0.05). **Creep:** no significant difference between groups (P>0.05).

Therefore, overall, myotonometer findings revealed the following: (a) Being performers of a sports practice that requires high habitual loading, traceurs' TA and MG muscles displayed the highest tone, stiffness and lowest relaxation time and creep, with their AT showing the highest stiffness and elasticity. (b) Being performers of a sports practice that requires moderate habitual loading, basketball players' TA and MG muscles displayed moderate tone, stiffness and relaxation time and creep characteristics and their AT stiffness and elasticity was also moderate. (c) Being people that practice only daily activities that require low habitual loading, non-athletes' TA and MG muscles displayed the lowest tone, stiffness and highest relaxation time and creep, and their AT was the least stiff and elastic.

4. DISCUSSION

The foot placement technique, sport specific habitual training and their interaction during drop landing were the factors examined in this study. The main findings were: (a) The major determinant was the foot placement technique for observed metrics, (b) FFL is more advantageous than THL in shock absorption, independently of training history, (c) sports specific habitual training, especially in landing technique, altered kinematic and neuromuscular strategies between groups to attain similar shock absorption and (d) habitual loading and training in sports specific landing technique altered mechanical characteristics of observed muscles and AT, in favor of the requirements of particular sports practice.

The foot and ankle initially encounter ground reaction forces and they are the first major structures to attenuate shock. Therefore, a minor difference in foot placement technique, heel contact being the only variable, changed GRF substantially in all groups. All GRF metrics of interest; pVF, time to pVF and LR showed that peak ground reaction forces were decreased during FFL, compared to THL. These results are in concert with the previous studies on FFL [50,70,72,75]. Our findings confirmed that FFL is more advantageous than THL in shock absorption, independently of training history and preferred landing technique. Furthermore, the difference in LR between traceurs and non-athletes for THL disappeared when performing FFL. Therefore, the effectiveness of FFL in shock absorption compared to the THL, was even more evident in non-athletes who had no previous FFL training.

First, the basic mechanical difference between the foot placement techniques should be examined to understand the distinct shock attenuation capacities. During FFL, first point of contact with the ground, forefoot, acts as the fulcrum of a type II lever throughout landing [109,116]. This creates an additional lever mechanism for FFL that initially counters GRF, whereas pivot point of the foot is closer to ankle for THL because of heel contact [117]. Type II lever is mechanically advantageous;

moment arm of the triceps surae and AT unit is longer than the moment arm of the body mass, that acts on tibia to talus. Due to this lever mechanism, muscle moments can counter loads with relatively less force. Furthermore, muscle-tendon forces are even higher for FFL compared to THL [10,118]. This is the result of elevated plantar flexor activation which leads to increased energy dissipation at ankle joint and force transmission through muscle-tendon complex [10, 70, 119]. Our findings of elevated GM EMG amplitudes for FFL also supports this notion. Presence of a mechanically advantageous lever mechanism and increased contribution of plantar flexors, together, increase shock attenuation in foot and ankle during FFL.

Throughout FFL, ground reaction forces were transmitted from forefoot through foot and ankle joint to lower leg. Note that windlass mechanism prevents longitudinal arch to collapse during this process and assists force transmission by stiffening the plantar fascia and increasing the activation of intrinsic and extrinsic muscles [119,120]. The direction of GRF relative to the tibia at the ankle joint affects the ratio of axial forces transmitted through tibia. When GRF transmitted from forefoot during FFL, the knee joint moments and tibial axial forces decrease, especially at the time of pVF compared to THL [89,90,121,122]. Furthermore, increased force transmission through triceps surae and AT unit could be directing GRF transmission posterior to the tibia during FFL, further decreasing tibial axial forces [89,90]. During THL, however, there is considerably less GRF transmission from forefoot. Majority of the BW collides with the ground at midfoot and heel and thus greater percentage of GRF acts on these site [123, 124]. Most of the GRF was directly transmitted through non-contractile tissues which leads to higher tibial axial forces for THL [70, 90]. Therefore, there is significantly less contribution of foot and ankle in shock attenuation during THL. This difference between techniques may further increase the advantage of FFL over THL in shock attenuation.

Joint kinematics and muscle activations also revealed between technique differences in all groups. However, these between technique differences were not at the same extent for each group. Ankle and knee were more plantar flexed and flexed, respectively, at IC for FFL. These joint positions at IC lead to reduced pVF, LR and anterior cruciate ligament (ACL) injury risk [10, 87, 88, 102, 125–127]. Increased plantar flexion at IC enables plantar flexors more time to contract over a wider ROM which lowers dorsiflexion velocities, as documented in this study, and prevents heel from touching the ground. This mechanism, also, may increase ankle joint's contribution to energy dissipation [70, 88, 125, 128].

No significant difference was found between groups for ankle, knee and hip joint angles at IC. Therefore, sport specific training did not affect joint angle regulation at IC. The adjustment of ankle and knee joint angles in preparation to FFL may be intrinsically programmed in all participants to increase plantar flexors' activation and bring them closer to their optimum length to maximize their force production and contribution in energy dissipation and stabilization [9, 88, 109, 129]. Furthermore, higher knee flexion at IC may be adjusted to move COM posterior during FFL to maintain balance, considering COM moves anterior with forefoot strategy [76, 130].

The ingrained mechanism for shock absorption in ankle starts before IC by pre-activation of plantar and dorsiflexors. In this study, TA and GM activation onset was between 90-110 ms and 115-125 ms, respectively. TA onset activation times were similar in previous studies that used same onset detection method (∼105 ms) [33,131]. GM activation onset was lower, at around 100 ms for landing from 52 cm in a previous study [132]. Higher landing height may be the reason for increased GM onset times of this study (∼120 ms). No significant difference was found between techniques. Activation onset is programmed by CNS, based on sensory input and predicted impact forces [34, 133]. Even though prediction of GRF is related to previous experiences, there was no significant difference between groups [131]. The participants that are not experienced in landing techniques may have learned and adjusted their feedforward mechanism intrinsically with just practice landings before trials.

The rate of EMG build-up after activation onset may differ between individuals, drop heights, landing surfaces and techniques [8, 34, 35]. In this study we found higher GM pre-activation levels for FFL in all groups, similar to findings of Kovacs et al. [70]. This indicates larger build-up for FFL, considering activation onset was similar between techniques. Higher GM pre-activation leads to increased voluntary motor contribution to post activation [35]. This contribution is especially crucial to control dorsiflexion velocities until SLRA of plantar flexors occurs after IC. In this study, SLRA of GM peaked at ~53 and ~52 ms after IC for THL and FFL, respectively. These times are very similar to previous findings (~53 ms) [26]. For THL, the vertical GRF peaked approximately 11 ms before SLRA in this study, which could be risky. Considering peak ACL strains and, thus, injuries occur around these times in sports that frequently perform THL, importance of pre-activation and its contribution to early motor control becomes evident [92, 100, 110, 134]. As for FFL, the vertical GRF was peaked approximately 17 ms after SLRA. Later occurrence of pVF reduces injury risk by allowing sufficient time for GM SLRA and AT moments (Electromechanical delay: ~12 ms) to respond pVF [111]. Moreover, higher GM activation during FFL may reduce ACL injury risk by lowering forces on the ligament [135].

Pre-activation levels may also affect short latency reflex amplitudes by modulating muscle spindle sensitivity [35, 136]. However, GM post 35-80 amplitudes (during which SLRA occurs) were different between techniques only in traceurs, even though pre-activation levels were higher for FFL in all groups. The relationship between angular velocity of ankle and SLRA may be the reason for these findings. Muscle stretch, thus ankle angular velocity is the primary factor for SLRA modulation [38, 114]. The difference in mean peak ankle velocity between techniques was more evident in traceurs (by ~88 °/sec) compared to basketball players (by ~14 °/sec) and non-athletes (by ~26 °/sec). Therefore, this difference in peak ankle velocity may have altered GM SLRA amplitudes between techniques in traceurs. Landings over 60 cm, SLRA starts to diminish despite of higher ankle angular velocities [114]. This assumed to be a protection mechanism to reduce muscle-tendon stiffness and, hence, extreme loads on those structures via increased inhibitory drive [137]. Decreased peak ankle velocity in traceurs with FFL may have diminished inhibitory mechanisms, leading to higher GM short latency reflex amplitudes.

The joint angles at the moment of pVF determine the magnitude and direction of peak GRF acting on musculoskeletal system and, also, muscles' capacity to respond it [89,90,103,138]. In this regard, later pVF time and, thus, higher knee and hip flexion angles at pVF may lower ACL loading and compressive loads on articular cartilage [92, 103, 127, 139, 140]. Therefore, FFL may lead to safer joint positions at pVF and lower injury risk.

Joint range of motions after pVF were higher for THL as expected, because of early pVF occurrence. Higher dorsi-flexion angles during THL were due to the requirement of the technique. As for group differences, higher dorsi-flexion in non-athletes compared to other groups may be the result of higher knee flexion in this group. Only anterior inclination of tibia via knee flexion may increase dorsi-flexion angle considering the heel is in contact with the ground during THL [129, 130].

Knee flexion after pVF may be increased to compensate for higher pVF during THL, which may reduce forces on ACL [91,141]. It must be mentioned that knee ROM did not differ between techniques, which indicates a sharper deceleration during THL after the occurrence of peak knee angular velocity, which is higher for THL.

Basketball players flexed their knee less than non-athletes after the time point of pVF. Knee extensor moments peak much after pVF because of neuromuscular response times and electromechanical delay [90, 92, 142–144]. Considering knee angle at pVF was not different between these two groups, difference in knee flexion between non-athletes and basketball players might be occurred with the involvement of knee extensor moments after pVF. Basketball players are expected to have higher knee extension strength and rate of force development than non-athletes, which could lead to faster deceleration after pVF [57, 145–147]. It should be noted that even though basketball players' faster deceleration during knee flexion may be beneficial for rapid movements of basketball, lower knee flexion coupled with high pVF and knee extensor moments can increase ACL loading and, thus, injury risk in those players [103,141,144,148–150].

Ankle dorsiflexion ROM was significantly higher for THL in traceurs and non-athletes, which was expected because of the requirement of the foot placement technique. However, dorsiflexion ROM did not differ between techniques in basketball players. Possible dorsiflexion limitation of basketball players may be the reason for similar dorsiflexion ROM between techniques [49]. Dorsiflexion limitation may not affect anterior inclination of tibia and, thus, knee flexion during FFL, due to already plantar flexed position. During THL, however, dorsiflexion limitation may decrease anterior inclination of tibia, considering the heel is in contact with the ground. This could lead to reduced ankle ROM during THL for basketball players, resulting similar ROM values for both techniques [105]. As a consequence, reduced dorsiflexion ROM in basketball players may lead to higher risk of patellar tendinopathy and ACL injury [14,151–153].

In previous studies greater dorsiflexion ROM resulted in lower GRF and higher knee flexion ROM [70,154]. In this study, however, higher dorsiflexion ROM during THL did not lead to lower GRF or increased knee flexion ROM. Therefore, foot placement technique (presence of heel contact) was the major determinant over dorsiflexion ROM for resultant GRF.

No difference was found between techniques in knee flexion ROM. Normally in THL knee joint's contribution to energy absorption is higher than ankle joint's [37,155]. FFL may decrease the ratio of energy dissipation in knee joint by increasing ankle joint's contribution [70,90,156]. This could potentially lower ACL injury risk by decreasing knee extensor moments and anterior tibial shear forces [90,103,155,157]. Moreover, higher post landing TA EMG amplitudes that were found in this study for THL, may further increase tibial anterior shear forces and, thus, ACL injury risk.

Basketball players flexed their knee less than non-athletes. It seems that non-athletes were collapsed more under load to decelerate and stop movement, which may indicate insufficiency in producing eccentric knee extension moments. Despite higher knee flexion ROM, pVF values were similar between non-athletes and basketball players. If non-athletes were asked to maintain similarly low knee flexion angles as basketball players, higher pVF values were to be expected, considering lower knee flexion results in higher pVF [23]. This may mean that non-athletes flexed more to compensate for similar ground reaction forces, which indicates an inability and strength deficiency for landing compared to basketball players. In a similar manner, studies comparing gender and athletic history showed that females and non-athletes flexed their knees more to dissipate GRF over a wider ROM compared to males and gymnasts, respectively [8, 37, 132]. This compensation mechanism with higher knee flexion ROM may reduce ACL loading, therefore ACL injury risk in non-athletes [14, 148, 150, 156, 158]. It should be noted that, however, it may also increase the risk of patellar tendinitis and patellofemoral pain syndrome after repetitive practice [159, 160].

Lower knee flexion ROM of basketball players may indicate higher leg stiffness during landing [105, 161, 162]. Basketball requires fast and frequent counter movement jumps, rapid acceleration right after landing, direction changes and reaching overhead by extending body for rebound, shooting and blocking. Therefore, higher stiffness in lower extremity may be beneficial for basketball players to augment power production and limit excessive flexion for quickness [78,104,107,117,163,164]. Possible restrictions in dorsiflexion ROM could be another reason for basketball players' lower knee flexion ROM compared to non-athletes [49, 105, 106, 165]. Limitations in dorsiflexion ROM may decrease knee flexion ROM by limiting anterior inclination of tibia [130]. These requirements of basketball may have led to basketball players' knee extensors to be stronger in more extended positions and hamstring/quadriceps strength ratios being higher than average [147, 166]. Due to these requirements and adaptations, basketball players might have developed a strategy which requires them to decelerate movement most effectively in a more extended knee position during landing [141]. Additionally, in the case of these strength adaptations, flexing too much may increase the injury risk of patellar tendon as muscles and tendon would not be prepared to counterbalance the GRF at those positions [159, 160]. It should be noted that, however, previous studies showed ACL loads and, thus injury risk was increased at low knee flexion angles [14,148]. Additionally, basketball players' longer extremities may further increase these loads due to longer moment arms of joints [167]. Therefore, restricting knee flexion ROM may be a risky strategy in the long run, even though it may be beneficial for performance.

The hip was flexed more during THL. Higher pVF and LR for THL may have led to higher hip flexion ROM in an effort to dissipate increased forces over larger ROM [8, 164]. Higher hip flexion angles may reduce pVF and ACL stress, therefore assumed to be safer [7, 14, 87, 150, 168], but it must be noted that GRF was found to be higher for THL. Therefore, higher hip flexion ROM for THL was probably a compensation mechanism in this study, not a safer strategy compared to FFL. FFL demonstrated lower pVF even with lower hip ROM.

Higher hip flexion ROM for THL may also be a consequence of posterior shift of COM after heel contact during THL [76]. Increasing hip flexion while knee angle is fixed results forward leaning of torso. Considering knee flexion ROM values were similar between techniques, leaning forward during THL might be an attempt to maintain balance by shifting COM from posterior to anterior.

Anterioposterior position shifts of COM may also explain between technique differences in TA and GM muscle activations. Muscles activate to keep the whole body stabilized after IC, besides their role in shock absorption [41, 169]. For THL, posterior shift of COM may lead to higher post TA EMG amplitudes to pull tibia and, thus, COM anterior in an attempt to maintain balance and prevent falling backward [170]. According to this assumption, increased activation of TA is expected to be evident after heel contact and following posterior shift of COM. The results of this study confirms this assumption. The heel contact and following impact occurs at the same time as pVF during THL, which is around 44 ms. In line with this, TA activation during THL became significantly higher than FFL during post 35-80 and post 80-200 in this study. Later occurrence of post max TA amplitude during THL supports this balancing function of TA.

As for FFL, COM is positioned posteriorly compared to THL [76]. This may lead to higher GM EMG amplitudes to pull the femur to the posterior in an attempt to balance COM over the a narrower base of support, forefoot, and prevent falling forward [170]. It should be noted that during FFL, contraction of GM along with other plantar flexors, also keeps heel from touching the ground at the distal end. Therefore, elevated activation of GM, especially during post 80-200, may be the consequence of these mechanisms for FFL.

The peak dorsiflexion velocity of ankle was found to be lower for FFL. The combination of elevated GM EMG amplitudes and relatively higher plantar flexion at IC to prevent heel contact, may have led to lower peak dorsiflexion velocity during FFL. The absence of this mechanism during THL may increase peak dorsiflexion velocity and, thus, impact force during heel contact, leading to higher pVF [34].

The ankle mean dorsiflexion velocity was lower for THL. The mean dorsiflexion velocity was calculated from IC to max dorsiflexion time. Max dorsiflexion occurred rapidly after IC for FFL, before ankle recoil. Therefore, mean angular velocity for FFL was found to be significantly higher than THL. It appears that while traceurs exhibited highest mean dorsiflexion velocity during THL, conversely they exhibited lowest mean dorsiflexion velocity for FFL. This distinction may indicate increased efficiency in control of ankle during landing with their habitual foot placement technique (FFL) and inefficiency with infrequent foot placement technique (THL).

The knee and hip peak and mean angular velocities were found to be higher for THL. Higher pVF and LR may have led to increased knee and hip angular velocities during THL compared to FFL [21,171–173]. Higher angular velocities may be beneficial for THL, due to large pVF and faster time to pVF. Joints encounter pVF at more flexed positions with increasing velocity, lessening loads on ACL [37, 92].

Co-activation index may provide important information about the regulation of joint stiffness [41]. In this study, the co-activation index was decreased throughout the landing, pre to post 0-35 and post-35-80 to post 80-200 being significant. In preparation to impact, as expected, pre co-activation was higher to stabilize the ankle joint in an attempt to enhance deceleration of dorsiflexion and prevent excessive joint angles [112, 174]. The co-activation continued to gradually decrease after IC, significantly dropping after 80 ms. In parallel to the decrease in co-activation, the activation of GM also gradually declined throughout the landing while activation of TA increased. This trend was the main reason for gradual drop in co-activation, because GM acts as antagonist and TA acts as agonist during dorsiflexion. Comparing techniques, co-activation index of THL was found to be higher during pre-activation, whereas co-activation index of FFL was higher during post 80-200. The differences between techniques is contradictory to other findings of EMG amplitudes and ankle angular velocities. Higher pre-activation of GM during FFL was expected to result in higher co-activation levels than THL, considering it acts as the antagonist. Furthermore, lower peak dorsiflexion velocities, also, during FFL indicate a stiffer ankle joint, hence higher-co-activation [174]. As for post 80-200, higher activation of TA during THL was expected to result in lower co-activation levels for THL. These contradictions in our findings may be the result of normalization method used in this study. Normalization with within trial peak EMG value may lead to inconsistencies for between technique comparisons, due to the difference in peak EMG of each technique in relation to its overall EMG.

Ankle deviation was found to be higher for FFL. Higher ankle deviation during FFL, considering ankle dorsiflexion ROM was higher for THL, may indicate the utilization of the tendon recoil mechanism in the ankle joint to a greater extent. During rapid energy-dissipating motions such as landing, tendon initially acts as an elastic spring by stretching rapidly after IC, and then shortening rather slowly, reducing the speed of muscle lengthening [175, 176]. This mechanism, however, is predominant during landings with heel contact, when the foot is stabilized on the ground. In the case of non-heel contact, as in FFL, the mechanism may change due to recoil of ankle. During initial recoil of ankle, the muscle-tendon unit may be acting more of a stiff spring for FFL than a damping mechanism as in THL. During movements that utilize stretch shortening cycle (SSC) such as drop jumps and hopping, tendon stores and then returns elastic energy, increasing movement efficiency and power amplification, while the muscles contract almost isometrically [108]. In the case of FFL, the ankle joint demonstrated a similar motion of a SSC during its recoil shortly after IC, even though the landing motion did not as a whole. Higher plantar flexion activation and AT forces during FFL support this assumption [10, 70]. Previous studies on recoil mechanism of AT during SSC, demonstrated higher plantar flexor activations, similar to our findings for FFL [109, 115, 177]. Longer and higher activation of plantar flexors during FFL may keep muscle fascicles close to their optimum lengths and increase their stiffness, enhancing the spring like behavior of AT [35,109,113–115]. During ankle recoil of FFL, the kinetic energy of initial mass, lower leg, may have been stored and returned before it was even transmitted to the forefoot and produced GRF [89]. This mechanism may lower resultant GRF and contribute to the energy dissipation in ankle joint, leaving knee and hip joints less work for shock attenuation [70, 109].

Lower ankle deviation of basketball players may be explained by the requirements of sport and specific adaptations to those requirements. Millett et al. reported higher ankle stiffness in netball players, a sport with similar movement requirements to basketball, compared to endurance athletes and control group [162]. Higher stiffness may be beneficial for the basketball performance [78,104,107,163]. As previously mentioned, possible restrictions in dorsiflexion ROM could be another reason for basketball players' lower ankle deviation as in dorsiflexion ROM [49, 105, 106, 165].

Sports specific habitual training and foot placement technique altered mechanical characteristics of lower leg muscles, TA and GM, and AT. The magnitude of these adaptations was relative to the loading requirements of particular sports practice. Traceurs frequently perform landing movements that requires to absorb relatively high loads. In line with sports' requirements, traceurs' TA and MG muscles displayed highest tone, stiffness and lowest relaxation time and creep, and their AT was the stiffest and the most elastic. These mechanical characteristics may be advantageous for parkour practice of traceurs [107, 178–182]. High habitual loading with repetitive practice of high impact landings, may have altered tissue characteristics via mechanical loading [60, 77, 81, 82, 180, 183–185]. Additionally, frequent practice of FFL technique may have contributed to the mechanical adaptations of tissue in traceurs. FFL increases forces on AT and GM, which further amplifies mechanical loading of tissue [10]. Following traceurs, basketball players' TA and MG muscles displayed moderate tone, stiffness and relaxation time and creep, and their AT stiffness and elasticity was, also, moderate among groups. Basketball practice requires landings with moderate loading, compared to other groups. Therefore, these findings are perfectly in line with the loading demands of the sports. As for non-athletes, their TA and MG muscles displayed lowest tone, stiffness and highest relaxation time and creep, and their AT was the least stiff and elastic, as expected. These findings suggested that sport specific habitual training and its loading demands altered mechanical characteristics of muscle and tendon, relative to the habitual loading intensity. These adaptations may lead to improvement in movement efficiency and safety, and thus athletic performance by enhancing force transmission and energy efficiency, relative to the requirements of sports practice.

5. CONCLUSION

The major determinant was the foot placement technique for landing mechanics. FFL technique found to be considerably advantageous than THL technique for shock absorption, independently of sport specific habitual training and preferred foot placement technique. All participants enhanced their shock attenuation capacity with the successful implementation of a simple verbal instruction on foot placement technique, "no heel contact". The effects of sport specific habitual training on landing mechanics became evident during implementation of the foot placement techniques. Kinematic and neuromuscular strategies were altered between groups to achieve similar shock attenuation for each technique. Participants adapted a suitable strategy, based on the potential adaptive effects of their sport specific habitual training, or no training, on their neuromuscular and musculoskeletal system. The adaptive effects of sport specific habitual training on musculoskeletal system were found out for mechanical characteristics of lower leg muscles (TA & GM) and Achilles tendon. Habitual training and concomitant loading patterns created advantageous mechanical adaptations in those structures for enhanced force transmission and elastic recoil capacity, relative to the requirements of particular sports practice. These adaptations may lead to improvement in movement efficiency and safety, and thus athletic performance. These findings present valuable insights into the effects of foot placement technique and sports specific habitual training on landing mechanics, as well as adaptive responses to sport specific habitual training in the mechanical characteristics of the lower leg muscles and Achilles tendon. Such insights may have practical implications for every practitioner and trainer whether in sports or recreational activities.

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