PASSIVE ANKLE-FOOT JOINT STIFFNESS

AND

PHYSICAL PERFORMANCE

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Passive Ankle-foot Joint Stiffness and Physical Performance

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CERTIFICATE OF ORIGINALITY

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MAN HOK SUM (Name of Student)

ABSTRACT

The toe flexor muscles maintain body balance during standing and provide push-off force during walking, running, and jumping. Additionally, they are important contributing structures to maintain normal foot function. Thus, weakness of these muscles may cause poor balance, inefficient locomotion and foot deformities. The quantification of metatarsophalangeal joints (MPJ) stiffness is valuable since it is considered as important factor in toe flexor muscles function. Ankle joint stiffness has been investigated for performance and clinical assessment. Previous researches suggested that eccentric exercise of ankle joint muscles increased its passive stiffness, hence improving running and jumping performance. Conversely, excessive ankle joint stiffness deteriorates ankle function. Measurement of MPJ and ankle joint stiffness is still largely depended on manual skills as current devices do not have good control on alignment, angular joint speed and displacement during measurement. Therefore, this study introduces an innovative dynamometer and protocol procedures for MPJ and ankle joint torque measurements with precise and reliable control of foot alignment, angular rotation speed and displacement. Within-day and between-day test-retest experiments on MPJ and ankle joint torque measurements were conducted on ten and nine healthy male subjects respectively. Intraclass-correlation coefficients (ICC) of averaged peak torque of both joints in within-day and between-day test-retest experiments were ranging from 0.91 to 0.96, and the joint torque was similar to the measurements of other studies. The results indicated the innovative device is systematic and reliable for the measurements and can be used for multiple scientific and clinical purposes. To investigate the relationship of ankle-foot

stiffness and performance outcome, 99 male subjects, aged between 18 to 30, were recruited for experiment. Passive MPJ and ankle joint torques were collected by the dynamometer. Physical outcome was evaluated by vertical stiffness (K_{vert}), which was determined by body deceleration and displacement of body mass during hopping on force platform. Pearson's correlation was analyzed. It was found that vertical stiffness was significantly correlated to passive ankle joint torque (T_{ankle}) and MPJ torques in both sitting and standing postures ($p<0.005$, $0.43>R>0.29$). To further examine the relationship of ankle-foot passive stiffness and physical performance, fifty-one out of 99 subjects were selected and divided into three groups according to their regular physical activities. There were 15 marathon runners, 19 basketball players and 17 miscellaneous group of athletes. It was hypothesized that the passive ankle joint torque of marathon runners is stronger because higher elastic energy can be stored and reused in stretching and shortening cycles during running for better running economy (RE). Secondly, it was hypothesized that passive MPJ torque of basketball players is stiffer because basketball players should run around the stage in different directions, sudden stop running, jump shooting and maintain the body balance. All these actions require strong toe flexors. One-way ANOVA was used to test between group differences. Post-hoc test showed that T_{ankle} of basketball players was significantly stiffer than that of miscellaneous athletes by about 24% ($p=0.03$). MPJ_{sit} and MPJ_{stand} of basketball players were stiffer than that of marathon runners and miscellaneous athletes by about 24% and 32% respectively ($p=0.01$). The results supported the second hypothesis that the toe plantar flexors of basketball players were stronger but not supported the first hypothesis that the ankle joints of marathon runners were stiffer for better RE. This study concluded that strengthening both ankle and toe plantar-flexors could improve basketball players' performance. In the last experiment, the Passive MPJ torque in sitting and standing position, total leg stiffness, vertical stiffness and RE during sub-maximal running were examined. It was found that RE, leg and vertical stiffness during sub-maximal running was correlated with passive MPJ torque. The improvement of RE can be explained by the reduction of contact time with increased toe strength, and hence improve leg and vertical stiffness. It was suggested that RE for sub-maximal running would be improved by increasing toe flexors strength. For further research, the dynamometer can be used to evaluate surgical outcome of toe deformity correction in term of range of motion, stiffness and maximum force output of the toes. Besides, it can be used as a conventional dynamometer to determine ankle joint stiffness for stroke patient to determine the range of motion and the force output of ankle and MPJ.

PUBLICATIONS ARISING FROM THE THESIS

1. Man, H.S., Leung, A.K.L., Cheung, J.T.M. and Sterzing, T., China Granted Patent Number CN 104013408 B. See appendix C.
2. Man, H.S., Leung, A.K.L., Cheung, J.T.M. and Sterzing, T., 2016. Reliability of metatarsophalangeal and ankle joint torque measurements by an innovative device. *Gait & Posture*, *48*, pp.189-193.
3. Man, H.S., Lam, W.K., Lee, J., Capio, C.M. and Leung, A.K.L., 2016. Is passive metatarsophalangeal joint stiffness related to leg stiffness, vertical stiffness and running economy during sub-maximal running? *Gait & Posture*, *49*, pp.303-308.
4. Man, H.S., Lam, W.K. and Leung, A.K.L., Ankle foot passive stiffness and vertical leg stiffness of runners, basketball and other sport players. Preparing to submit.

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Last but not least, I wish to express my sincere gratitude to Mr. Pete Wong, associate engineer of the Industrial Centre for his valuable recommendation on hardware design and fabrication of the innovative device.

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LIST OF ABBREVIATIONS

ADP	Adenosine diphosphate
ATP	Adenosine triphosphate
a_z	Acceleration of CoM in vertical direction
bpm	Beat per minute
BW	Body weight
CoM	Centre of mass
$F_{Z(max)}$	Maximum vertical force
ICC	Intraclass-correlation coefficients
K_{leg}	Leg stiffness
K_{vert}	Vertical stiffness
k	Stiffness of ideal spring
MPJ	Metatarsophalangeal joints
MPJ_{sit}	Passive MPJ torque in sitting position
MPJ_{stand}	Passive MPJ torque in standing position
P	Period of oscillation
RE	Running economy
ROM	Range of motion
RER	Respiratory exchange ratio
T_{ankle}	Passive ankle joint torque
t_a	Aerial time
t_c	Contact time
t_{ce}	Effective contact time
v	Horizontal velocity
Δ y	Displacement of CoM
Δ y(max)	Maximum displacement of CoM
Δ L	Change in leg length
v'	Hopping velocity
ω	Natural frequency of oscillation
SF	Stride frequency
SL	Stride length
L	Leg length

CHAPTER 1. INTRODUCTION

1.1 Ankle-foot stiffness - Background

Ankle passive stiffness is relatively well understood to cope with stroke in rehabilitation (Lamontagne et al., 2000, Vattanasilp et al., 2000, Kim et al., 2005, Selles et al., 2005, Kwah et al., 2012, Cho et al., 2013, Moon et al., 2013, Jung et al., 2015) and calf muscle training for sport performance (McNair et al., 2001, McNair et al., 2002, Bressel et al., 2004, Arampatzis et al., 2005, Araújo et al., 2011). Off-the-shelf automatic dynamometers such as Biodex are available to measure the ankle joint passive stiffness. However, to my best knowledge, only one custom made device was designed to measure the first metatarsophalangeal joint passive stiffness (Rao et al., 2011), and the dynamometer was manually manipulated in rotation speed and there was no aid for alignment control. The metatarsophalangeal joints (MPJ) connect the toes and other parts of the foot to provide body balance. They allow the toes plantarflexion and dorsiflexion, as well as shares the plantar pressure of the foot during stand phase and provides propulsion force during toe-off phase in running and walking (Mann and Hagy, 1979). Hence, the strength of the plantarflexor muscles of the toes are very important in agility and body balance. The passive extension stiffness of the toes partially comes from the tension of plantar aponeurosis and the extrinsic muscles of the toes. These structures maintain the normal foot arch which is believed to store and release elastic energy on each step for energy efficiency (Alexander et al., 1987). Investigating the passive stiffness and the biomechanical behavior of the MPJ is as worth as that of ankle joint.

1.2 Objectives of this study

The main objective of this study was to invent an innovative computer-controlled dynamometer for both the ankle joint and the MPJ to provide reliable passive joint torque measurement. Hence, the ankle-foot passive joint torque determined by the device was correlated functionally in terms of vertical and leg stiffness and eventually the performance outcome in term of running economy (RE).

1.3 Outline of the Dissertation

Chapter 1 introduces the background and objectives of present study, and provides summaries of each chapters.

Chapter 2 is the literature review on the general foot anatomy related to current study, the spring-mass model of the leg and the physiology of running economy. The functional and structural characteristics of dynamometers used to measure ankle-foot stiffness are summarized.

An innovative dynamometer for ankle-foot passive stiffness measurement is described in chapter 3. The device could measure passive torque of the ankle joint and the MPJ in different configurations. With-in day and between day test-retest experiments were conducted. The Intraclass-correlation coefficients (ICC) of the device was over 0.9.

Chapter 4 investigates the relationships of passive ankle joint and MPJ torques and vertical stiffness (K_{vert}) during hopping. The Pearson correlation indicated that K_{vert} was significant ($p<0.005$) but weakly ($0.29<R<0.43$) correlated to passive ankle and MPJ torques, The passive ankle and MPJ torques were moderately correlated ($p<0.001$, $0.41<R<0.63$).

Chapter 5 compares the ankle-foot stiffness of the marathon runners, basketball players and other athletes to examine how the nature of sport changed the mechanical properties of the human foot for adaptation. The results of One way ANOVA and post-hoc tests indicated that T_{ankle} of basketball players was significantly stiffer than that of miscellaneous athletes by about 24% ($p=0.03$). Besides, MPJ_{sit} and MPJ_{stand} of basketball players were stiffer than that of marathon runners and miscellaneous athletes by about 24% and 32% respectively ($p=0.01$). Strengthening both ankle and toe plantar-flexors could improve basketball players' performance.

Chapter 6 introduces an pilot test that investigated the relationship among Passive MPJ torque, energy efficiency, leg and vertical stiffness during sub-maximal running. Passive MPJ torque was first measured. The subjects were then put on wireless gas analyzer to collect breathing oxygen and carbon dioxide volume to determine the running economy during submaximal running. At the same time, pressure sensors insoles were inserted to the shoes to collect information of peak force, contact time and flight time of each step to determine the K_{leg} and K_{vert} of each step. It was found that passive MPJ torque was correlated with RE, leg and vertical stiffness during sub-maximal running. The results supported the hypothesis that strong toe flexors reduce the contact time by improving the body balance during stance phase. The shortened contact time led to increase of both K_{leg} and K_{vert}, which in turn improveed RE. This study demonstrated that RE for sub-maximal running would be improved by increasing toe flexors strength.

Chapter 7 is the conclusion. The key findings of experiments, such as the high reliability of the custom-made device, the passive MPJ torque difference of basketball players, runner and miscellaneous group of athletes, and the high correlations of passive MPJ torque with K_{leg} and K_{vert} and running economy during sub-maximal running are summarized. Limitations of present study such as the subject selection criteria and the direction for future research are included in this chapter.

CHAPTER 2. LITERATURE REVIEW

2.1 The Foot

The human ankle-foot complex is so complicated that it comprises of 30 bones and 33 joints which are connected by more than hundred muscles, tendons and ligaments. Figure 2.1 shows the overview of foot bones.

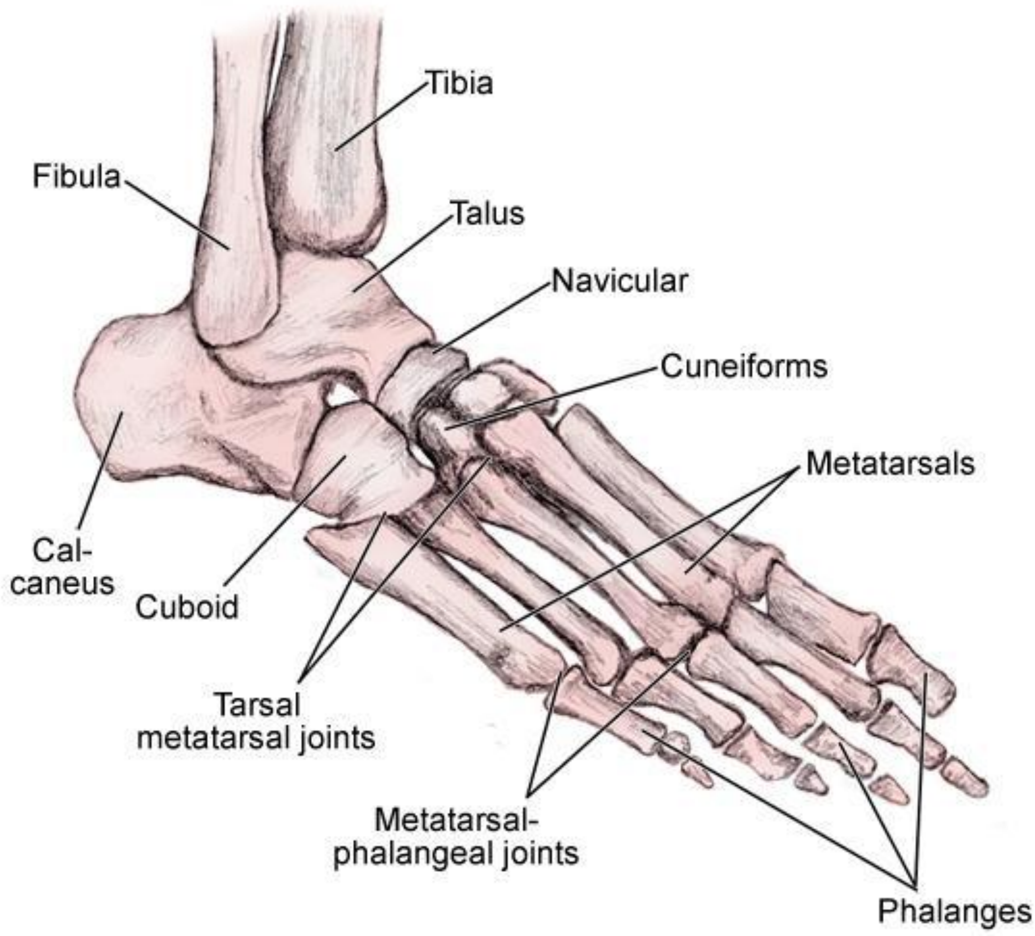


Figure 2-1 Overview of human foot bone.
(Source: http://emedicine.medscape.com/article/1922965-overview)

The foot is commonly divided into three parts: hindfoot, midfoot and forefoot. The hind foot which consists of the talus and calcaneus, is the proximal part of the foot. The midfoot is composed of the cuboid, navicular and three cuniforms. The forefoot is composed of the metatarsals and phalanges. There are two sesamoid bones under the first metatarsal. The main function of the foot includes body weight support without high RE in standing. It also works as a rigid lever to push the body forward in walking and running. In addition, the multiple joint structure and the soft tissue on the plantar side allow the foot to absorb part of the shock in movement on either flat surface or uneven terrain. In this thesis, only the structures related to the ankle joint and MPJ are reviewed to focus on the topic.

2.1.1 Ankle

The ankle joint, or talocrural joint, is the articulation between the head of the talus, distal tibia and distal fibular. It is a synovial joint which primarily acts as a hinge joint to provide dorsiflexion and plantarflexion of the foot. The bones are mainly linked by the anterior talofibuar ligament, talofibular ligament, calcaneofibular ligament, and posterior talofibular ligament originated from the fibula, as well as the anterior talotatar ligament, posterior talotatar ligament and tibiocalcaneal ligament originated from the tibia as shown in Figure 2-2.

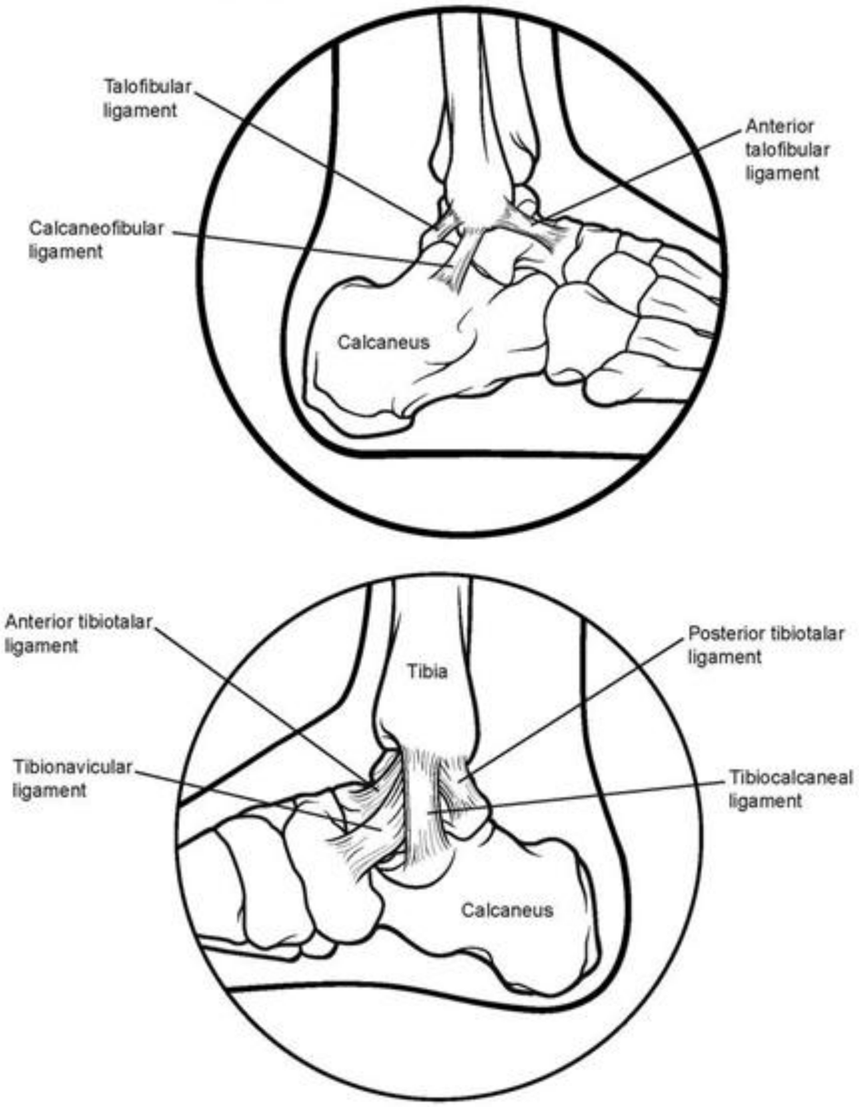


Figure 2-2 The main ligaments that hold tibia,fibula, talus and calcaneum (Source:http://emedicine.medscape.com/article/1922965-overview)

The proximal part of the ankle joint is composed of the distal tibia and the malleoli of the tibia and fibula to form a concave surface. The distal articulation surface is formed by the body of the talus, which has three articular surfaces. The horse horn structure of the ankle joint allows the foot to have about 20° of dorsiflexion and 50° of plantar flexion.

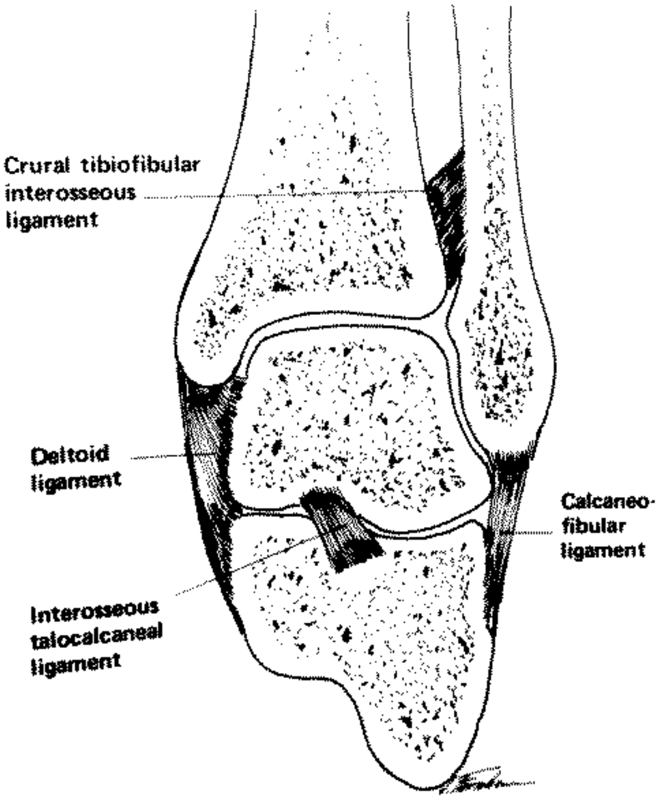


Figure 2-3 The cross section of ankle to show the horse-horn talus joint structure articulated with tibia and fibula (Source: Joint structure and function, a comprehensive analysis, second edition, page 389)

The main movement of the ankle joint is plantar flexion. The horse-horn talus joint structure articulated with tibia and fibula as shown in Figure 2-3. The motion is achieved by the strong gastrocnemius, soleus and plantaris muscles. In walking and running, the muscles contract to lift the heel and throw the body to move forward by using the foot as a lever. In swing phase, the plantar flexors relax and the dorisflexor muscles contract and dorsiflex the foot to avoid the foot hitting the ground.

Although there are many ligaments connecting the ankle joint, the stiffness of the ankle joint is mainly contributed by the Achilles tendon which connects the plantaris, gastrocnemius (calf) and soleus muscles. In stroke patients the activities of the lower limb muscles are disrupted by the absence of nerve impulse from the brain. The muscles without stimulation will become atrophic. Since the strength of dorsiflexors is weaker than the plantarflexors in ankle joint, the atrophy of the counteracting muscles will eventually keep the foot in a plantar flexion position. It is known as the drop foot symptom.

2.1.2 Metatarsophalangeal Joints (MPJ)

The MPJ refers to the condyloid synovial joint connecting the metatarsal and proximal phalanges. The MPJ consist of 5 metatarsophalangeal joints (1^{st} to 5^{th}) in one foot. The metatarsals and the proximal phalanges are linked by the plantar ligaments, and deep transverse metatarsal ligament and the transverse head of the adductor hallucis as shown in Figure 2-4. These joints can provide 82°of extension and 17°of flexion (Buell et al., 1988). The toe extensors include the extensor hallucis longus, extensor digitorum longus, and extensor digitorum brevis. The toe flexors include the flexor digitorum brevis, lumbricals, interossei, flexor hallicus brevis, flexor digitorum brevis, flexor hallucis longus, flexor digiti minimi brevis, flexor digitorum longus, and quadratus plantae. The flexor hallucis longus and the flexor digitorum longus (Figure 2-5) are powerful toes plantar flexors that stabilize the foot at mid stance and provide propulsion at toe off.

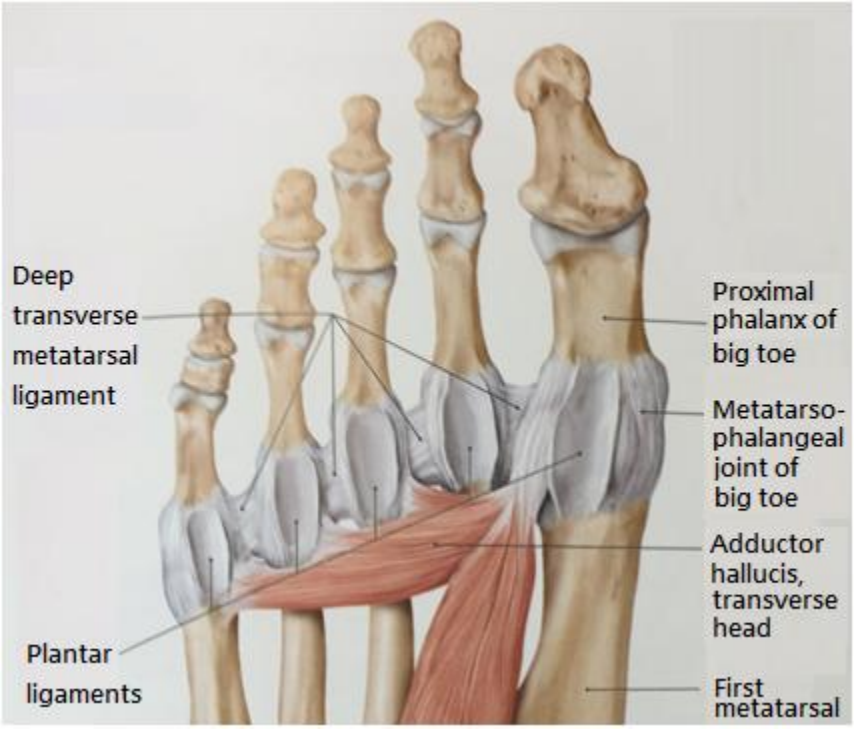


Figure 2-4 The ligaments and muscles that link MPJ of the five toes (Source: Thieme, Atlas of Anatomy, page 439)

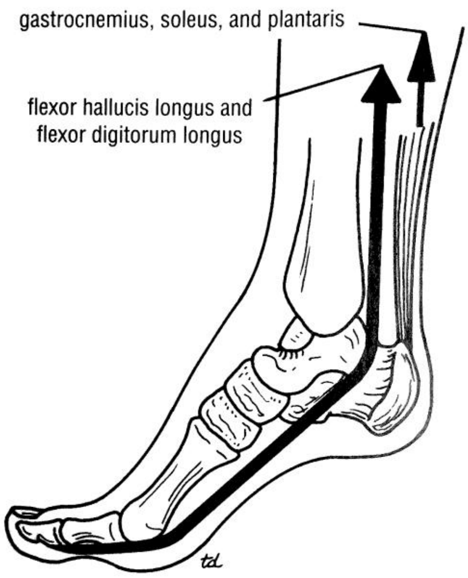


Figure 2-5 Toe pushing the ground by the contraction of flexor hallucis longus and flexor digitorum longus
(Source: Clinical Anatomy for medical student, 5th edition, page 588)

Apart from plantarflexion-dorsiflexion, the toes can perform adduction and abduction. The abductor hallucis is the abductor of the first toe. Its origin is located on tuberosity of the calcaneus and it inserts on medial side of the first phalanx of hallux. The interossei dorsal muscles are adjacent to the sides of metatarsal bones and insert to the base of the proximal phalanges to provide abduction at the metatarsophalangeal joints of the third and fourth toes. The adductor hallucis is the big toe adductors. Its insertion is located on the lateral side of the base of the first phalanx of the first toe and the sesamoid bones. The adductor has oblique head which originates to the proximal ends of the middle 3 metatarsal bones and the other transverse head originates to the metatarsophalangeal ligaments of the lateral 3 toes. The interossei plantar muscles are toe adductors of the third to the fifth toes which originate from the inferior surfaces of the third to fifth metatarsal bones and insert to the medial side of the bases of the proximal phalanges of the three lateral toes.

2.1.3 Plantar aponeurosis

The plantar aponeurosis, or plantar fascia, is a thick and rigid fascia that arises from the medial and lateral tubercles of the calcaneus. The fascia divides at the toes into five slips, and hence covers most of the plantar side of foot. The slips eventually attach to the proximal phalanx of each toe via the deep transverse metatarsal ligament and the fibrous flexor sheath of foot. When the toes are extended at the MPJ, the fascia is pulled and becomes tightened. The metatarsal heads perform as a pulleys. Eventually, the foot becomes shorter but higher. This is called the windlass effect. The tightened fascia resists excessive toe tension by creating a passive flexor force across the MPJ during stand phase. In addition, the plantar aponeurosis provides stability to the first metatarsophalangeal joint (Doty and Coughlin, 2013).

2.1.4 Foot arches

The medial longitudinal arch, lateral longitudinal arch and transverse arch maintain the foot shape. The foot arches are supported by different mechanisms. The wedge shape of the talus acts as a 'keystone' to support the medial longitudinal arch. The plantar aponeurosis acts as a tie beam and the peroneus longus provides a suspension bridge mechanism to support the arch. Among the 3 arches the medial longitudinal arch is the highest. The arches act as springs to allow stretch and store elastic energy in the first half of the stance phase. They become release in the second half for energy saving (Alexander et al., 1987).

2.1.5 Foot morphology

The normal foot consists of healthy structure of bones and soft tissue. Tension imbalance of ligaments and tendons of the foot may cause toe deformities such as hammer toes, claw toes (Myerson and Shereff, 1989) and hallux valgus (Doty and Coughlin, 2013). In addition, foot morphology is influenced by varieties such as age and gender (Tomassoni et al., 2014).

2.2 Joint stiffness

Joint stiffness, or sometimes called torsional stiffness, refers to the rotational resistance of individual joint. It is the ratio of joint torque to angular displacement. Active joint stiffness and passive joint stiffness are the measures of particular joint torque in motion and in relaxation respectively.

2.2.1 Active joint stiffness

Joint stiffness of the hip, knee and ankle are generally determined by a technique called inverse dynamics. This is formulated by the kinematics of body segment and force data collected by 3D motion capture system and force platform respectively. Lower limb segments of the thigh, leg and foot are distinguished by joint markers placed on bony landmarks such as the shank, greater trochanter, lateral epicondyle of femur, lateral malleolus of ankle, 5th metatarsophalangeal joint and tip of big toe. Linked segment model is established according to the position of the markers collected by video cameras. In addition, ground reaction force is recorded by force platforms when the foot is stepping on it (Figure 2-6). Eventually, joint moment can be determined by inverse dynamics according to the kinematic and kinetic data (Figure 2-7).

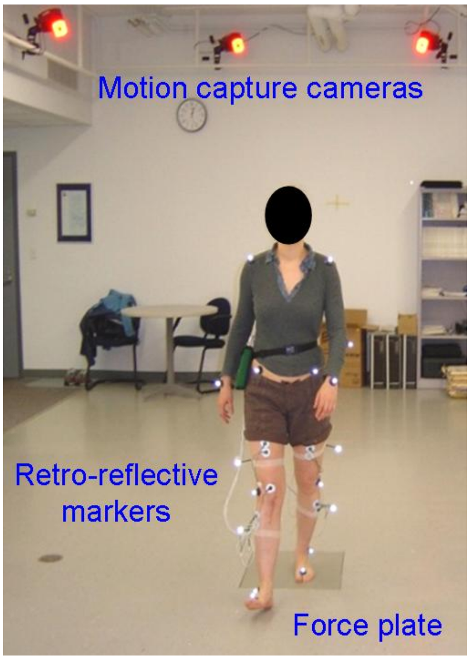


Figure 2-6 Retro-reflective markers are utilized to distinguish body segments, and the subject is walking on force platform to collect kinematic and kinetic data.
(Source: https://thebiomechanist.wordpress.com/ category/biomechanics/)

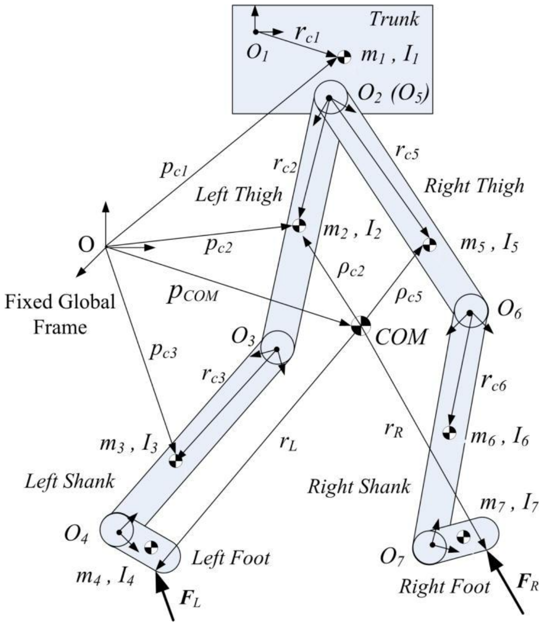


Figure 2-7 Joint moment can be determined by inverse dynamics according to kinematic and kinetic data. (Source:http://biomechanical.asmedigitalcollection.asme.org/article.aspx?articleid=1476110)

2.2.2 Passive joint stiffness

Passive stiffness of joint is measured by isokinetic dynamometer in relaxation (Figure 2-8). A limb segment is first fixed to a dynamometer and with the target joint axis aligns with the rotating axis of the device. Then the joint is taken through a range of motion passively. The torque resistance is measured during the movement, and the passive stiffness of joint is regarded as the ratio of passive resistance torque to angular displacement.

Figure 2-8 Joint passive stiffness can be determined by isokinetic dynamometer
(Source:http://www.lookfordiagnosis.com/mesh_info.php?term=Muscle+Strength+Dynamometer&lang=1)

2.2.3 Review of ankle joint dynamometers

Computer-controlled dynamometers, such as biodex and cybex, are designed for multiple joints stiffness measurement. The servo-motor of the dynamometer applies repeatable angular speed and displacement for passive joint movement. Apart from plantar flexion and doris flexion, these devices can also allow eversion and inversion for measurement of joint stiffness. An example, Biodex is shown in Figure 2-9. However, these devices are not appropriate for some clinical setting due to their large size.

Figure 2-9 Biodex dynamometer, measurement of ankle stiffness joint in plantar-doris flexion and eversion-inversion movement.
(Source: http://www.biodex.com/physical-medicine/products/dynamometers/system-4-pro)

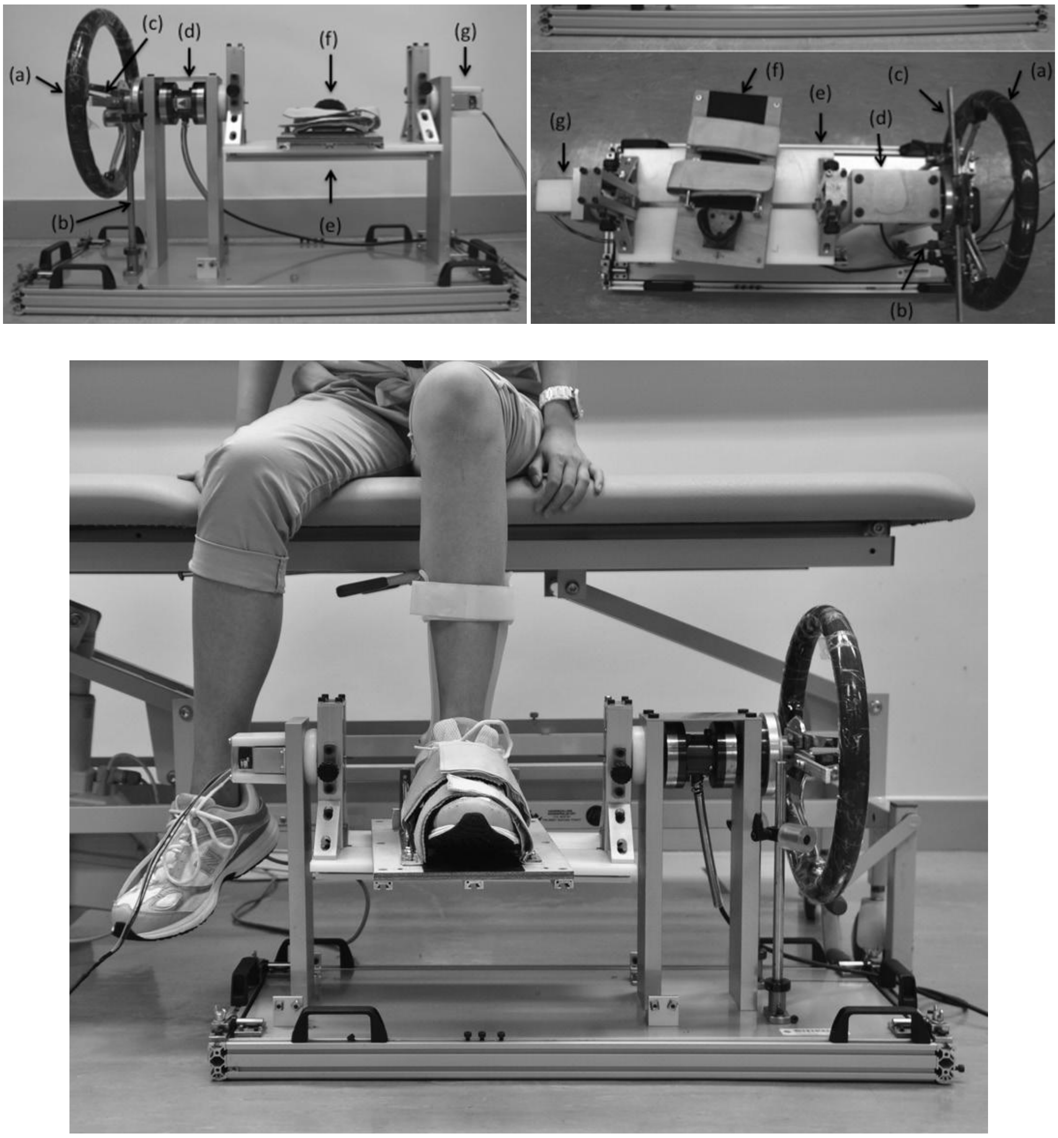


Figure 2-10 A manually controlled custom-made ankle joint dynamometer .
(Source: Kobayashi et. al (2011), Design of a manual device to measure ankle joint stiffness and range of motion)

Figure 2-10 shows a custom-made dynamometer for passive ankle joint plantar flexion and doris flexion stiffness measurement (Koboyashi et al., 2011). The upper figure indicates (a) steering wheel; (b) mechanical stopper; (c) bar; (d) torquemeter; (e) rotary plate; (f) foot plate; (g) potentiometer. The lower figure shows measurement of ankle joint stiffness of a subject wearing shoes and ankle-foot orthosis. This manually-controlled device saves equipment expense and facilitates clinical use. However, the absence of computer control motor leads to some sort of inaccuracy since the user may not control the rotation speed precisely enough.

Thirty out of thirty-four published articles adopted computer- controlled dynamometers to measure ankle joint passive stiffness (Table 2-1). Nakamura et al. (2011, 2012) did not indicate whether automatic or manually-controlled dynamometers was used. Only Kobayashi et al., (2011) used manually-controlled dynamometer to measure ankle joint stiffness. Most of the scientific research adopted automatic-driven dynamometer to measure ankle joint passive stiffness because this kind of device could provide constant and fine-adjustable angular speed and angular displacement. However, the dynamometers employed in previous researches did not accurately control the alignment between the device and the tested limb segment. It is difficult to define the range of ankle stiffness based on these previous researches even though the angular speed and angular displacement could be accurately controlled. The subject groups of previous experiment could be distinguished as normal subject group (Amin and Herrington, 2014, Araújo et al., 2011, Gajdosik et al., 2007, Hopper et al., 2014, Konrad et al., 2015, Lamontagne et al., 1997a, Maïsetti et al., 2012, Malmir et al., 2014, McNair et al., 2001, McNair et al., 2002, Nakamura et al., 2011, Nakamura et al., 2012, Porter et al., 2002, Sobolewski et al., 2013, Sobolewski et al., 2014, Tilp et al., 2012, Whitting et al., 2013), post- Achilles tendon injured group (Bressel and McNair, 2001, Bressel et al., 2004, McNair et al., 2013) and patients with central nervous disorder (Boiteau et al., 1995, Jung et al., 2015, Kim et al., 2005, Kobayashi et al., 2011, Lamontagne et al., 1997b, Lamontagne et al., 2000, Marsden et al., 2012, Matthiasdottir et al., 2014, Moon et al., 2013, Ross and Engsberg, 2002, Singer et al., 2008). Moreover, there was no clear guideline for the posture of measurement. The majority of the reviewed articles preferred to take measurement in seating position with either the knee fully extended (Kemertzis et al., 2008, Lamontagne et al., 1997a, Lamontagne et al., 2000, Maïsetti et al., 2012, Porter et al., 2002, Sobolewski et al., 2014, Tilp et al., 2012) or partially flexed (Kobayashi et al., 2011, Lamontagne et al., 1997b, Matthiasdottir et al., 2014, McNair et al., 2013, Moon et al., 2013, Ross and Engsberg, 2002), but some other studies selected prone lying (Nakamura et al., 2011, Whitting et al., 2013) or supine lying (Araújo et al., 2011, Bressel et al., 2004, McNair et al., 2001, McNair et al., 2002, Singer et al., 2008) position. More importantly, rotation speed and displacement of ankle were widely different in previous experiments. For angular displacement, some experiments kept the doriflexion-plantarflexion range in about 10 to 20° to stimulate the range of motion in normal walking (Kim et al., 2005, Mahieu et al., 2008). However, some other experiments required the subject passively dorisflex the ankle until uncomfortable (Bressel et al., 2004, Sobolewski et al., 2013). The ankle joint stiffness obviously increase with ankle joint dorsiflexion due to the progressive elongation of Achilles tendons. The range of angular speed was even wider in previous studies, and started from 5°/s (Araújo et al., 2011) to 190°/s (Boiteau et al., 1995). Since the passive stiffness of ankle depends on angular speed (McNair et al., 2002), it is difficult to conclude the ankle joint stiffness according to previous studies. Despite the different protocols of experiment, it is still possible to summarize the ankle joint passive stiffness in different conditions for reference. McNair et al., (2013) found that the ankle joint passive stiffness was about 15Nm in the normal side and 10 Nm in the injured side of subject with post ruptured side of Achilles tendon in a rotational speed of 5°/s and 80% of the maximum range of motion in sitting position. For hemiplegic patients, the averaged ankle joint stiffness was 5 Nm and 15 Nm in angular speed of 5°/s and 25°/s respectively when the foot was passively dorsi-flexed to 80% of the maximal dorsiflexion range in supine lying position (Singer et al., 2008).

Apart from plantar-flexion and dorsiflexion, devices for inversion of ankle joint were invented in laboratories. Forty-six journal articles were published during 1981 to 2012 (Ha et al., 2015), and the invented devices were classified as tilt platforms, trapdoors, and dynamic fulcrum devices. However, these devices were not designed for joint torque measurement but stimulated the environment for ankle supination sprains. Since this thesis investigated joint stiffness and physical performance rather than the mechanism of ankle joint injuries, the ankle inversion sprain simulators were out of the scope.

Table 2-1Summary of recent researches that related to dynamometers used to measure ankle joint passive stiffness.

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Amin and Herrington, 2014)	Ankle joint passive stiffness and balance with single leg	Measure CoM of body in single legged standing	Plantarflexion stiffness was significantly associated with better control of balance in unilateral stance amongst athletic adults	21 healthy, adult athletes (12 men and 9 women)	KinCom dynamometer
(Araújo et al., 2011)	Validity and reliability of clinical tests for assessing passive ankle stiffness	Intra- and inter-examiner reliability test by two examiners	ICC=-0.81 to -0.88 ($p<0.001$) for the correlation between the passive ankle stiffness	15 healthy subjects (seven men and eight women)	Biodex dynamometer
(Boiteau et al., 1995)	Test-retest reliability	Compare the test-retest reliability of two methods	High ICCs for the resistive force	10 children with spasticity	hand-held and Kin-Com dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Bressel and McNair, 2001)	Ankle joint angle, passive torque, and maximal isometric plantar flexor torque for subjects who had ruptured their Achilles tendon	Measure ankle joint stiffness after an Achilles tendon rupture	Isometric torque and peak passive torque were 17% and 10% greater for the uninvolved versus the involved limb	40 people (26 men and 14 women) with a repaired Achilles tendon rupture	KinCom dynamometer
(Bressel et al., 2004)	Achilles tendon rupture on the proprioception and kinetic performance, ankle joint angle and passive torque	Measure ankle joint proprioception and passive stiffness	Participants with a previous history of an Achilles tendon rupture display proprioception deficits in both limbs and greater torque	20 subjects with Achilles tendon rupture and 20 normal subjects	Biodex dynamometer
(Gajdosik et al., 2007)	Adaptation of calf muscle-tendon unit after stretching exercise	Passively stretched calf muscle-tendon unit	Maximal passive dorsi-flexion angle, torque and range of motion increase, passive stiffness did not change	6 women with stretching exercise and 4 women in control group	Kin-Com dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Hopper et al., 2014)	The effect of ankle taping on plantar-flexion strength, angle matching and force matching.	Apply ankle taping in different method for comparison	Subjects were most accurate at matching a plantar-flexion angle of 20	20 healthy women	Dual Ankle Dynamometer (DAD) (Ribuck Industries, Western Australia)
(Jung et al., 2015)	Compare the reliability of Biodex and hand-held goniometer	By reliability test	Biodex dynamometer had better intra- and inter-reliability measurements	3 female and 12 male stroke patients	Biodex and hand-held goniometer
(Kemertzis et al., 2008)	Voluntary ROM, peak torque, and corresponding joint angle of the plantar- and dorsiflexors	Measure ankle torque after whole-body vibration	Shift in the angle of peak plantarflexor torque production corresponding to a longer muscle length	20 healthy men	Biodex dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Kim et al., 2005)	Peak torque, work, and threshold angle in passive movements of the ankle	Electromyography of post-stroke ankle plantar flexor spasticity	Peak torque, threshold angle, work, and rectified integrated electromyographic activity were significantly higher in the post-stroke spastic group	20 post-stroke hemiplegic patients and 10 normal subjects	Cybex dynamometer
(Kobayashi et al., 2011)	Ankle joint stiffness and ROM	Fix the foot on the pedal and passively rotate the wheel in different trials	High ICC	Ten male subjects with hemiplegia	Manual custom-made device
(Konrad et al., 2015)	Various parameters of the human gastrocnemius medialis muscle and the Achilles tendon.	Measurement of ankle joint stiffness of pre and post stretching training program	Mean range of motion increase, tendon stiffness decrease	49 healthy subjects, 25 in stretching group and 24 in control group	CON-TREX MJ, CMV AG, Duebendorf Switzerland
(Lamontagne et al., 1997a)	Viscoelastic behavior of ankle	Passively stretched calf muscle-tendon unit in different speed	Resistive toque responses to increasing movement velocity	18 healthy subjects, 6 men and 12 women	Kin-Com dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Lamontagne et al., 1997b)	Impaired viscoelastic behavior of spastic plantar-flexors during passive stretching at different velocities	Passive ankle dorsi-flexions of 6 chronic (1-3 yr) spinal cord injuries were tested	Torque resistance was significantly higher than normal group	6 spinal cord lesions (C6-T10) patients and 12 healthy controls	Kin-Com dynamometer
(Lamontagne et al., 2000)	Passive stiffness of the ankle plantarflexor moment	Compare paretic side and non-paretic side	On the paretic side, passive stiffness contributed more to total plantarflexor stiffness during gait compared with both the nonparetic side and control values	14 stroke patients and 11 healthy subjects	KinCom dynamometer
(Maïsetti et al., 2012)	Characterization of passive elastic properties of the human medial gastrocnemius muscle belly	Correlate shearing modulus of gastrocnemius using super sonic shear imaging and dynamometer	High correlation	7 sedentary healthy men	Biodex dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Mahieu et al., 2008)	The mechanical properties of the plantar flexors muscle-tendon tissue	Use dynmomemeter and ultrasonic probe	The eccentric heel drop program also resulted in a significant decrease of the passive resistive torque of the plantar flexors	37 subjects with eccentric training and 37 subjects in control group	Biodex dynamometer
(Malmir et al., 2014)	Viscoelastic response of the lateral side of the ankle to cyclic movement	Measure ankle stiffness in eversion	Significant difference between the means of energy absorption for the first 20th repetitions	18 recreationally active healthy males	Biodex dynamometer
(Marsden et al., 2012)	Passive stiffness and spasticity was assessed during motor-driven slow (5°/s) and fast (60°/s) stretches at the ankle	Motor-driven slow (5°/s) and fast (60°/s) stretches at the ankle	Passive stiffness, assessed during slow stretches, was 35% higher in the plantarflexors in people with spasticity	20 patients with spasticity and 18 matched controls	Biodex dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Matthiasdottir et al., 2014)	Fascicle lengths in passive movement of the ankle joint	Ultrasound imaging was used to measure fascicle lengths while a dynamometer moved the ankle joint through the range of motion	Fascicle lengths in children with cerebral palsy were 43% smaller than those for control subjects throughout the range of motion	11 children with spastic cerebral palsy and 14 controls	Biodex dynamometer
(McNair et al., 2001)	Ankle stiffness and force relaxation response	Stretching at the ankle joint viscoelastic responses to holds and passive motion	Stiffness was decreased significantly ($P<0.05$) for the continuous passive motion condition only	24 healthy subjects (8 women and 15 men)	KinCom dynamometer
(McNair et al., 2002)	Peak passive force and average stiffness when ankle rotated in 5°/sec and 25°/sec	Ankle joint from 0° to 80% of maximum doris flexion over a 2 min period	Peak force was significantly higher at 25°/s for the first repetition	18 healthy subjects (11 women and 7 men)	KinCom dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(McNair et al., 2013)	Energy stored, stiffness, and shock absorption in the plantar flexor muscle – tendon unit	Measure ankle joint mechanical properties 6 months post-rupture of the Achilles tendon.	Passive torque in unaffected legs was greater (26%) than affected legs, energy stored in affected legs was 80% of that in unaffected legs	38 subjects with Achilles tendon rupture	Biodex dynamometer
(Moon et al., 2013)	Effect of extracorporeal shock wave therapy on ankle spasticity in minor stroke patients	Extracorporeal shock wave applied on lower limb spasticity in sub-acute stroke patients	Lower limb spasticity in sub-acute stroke patients was significantly improved immediately after extracorporeal shock wave therapy	30 subacute stroke patients (13 women and 17 men)	Biodex dynamometer
(Nakamura et al., 2011)	Muscle-tendon unit properties after 10 minutes static stretching	Static stretching on the passive stiffness of the human gastrocnemius muscle tendon unit	Muscle tendon unit stiffness and muscle stiffness significantly decreased at both immediately and 10 min after static stretching	15 healthy males	(Myoret, Kawasaki Heavy Ind., Kobe, Japan, ultrasonography

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Nakamura et al., 2012)	ROM, passive torque, myotendinous junction displacement and, muscle fascicle length of the gastrocnemius muscle	Pre and post 4-week static stretch training program	ROM and myotendinous junction displacement significantly increased, and the passive torque significantly decreased	18 healthy males	MYORET RZ-450; Kawasaki Heavy Industries, Kobe, Japan
(Porter et al., 2002)	Stiffness after eccentric exercised	Unaccustomed eccentric exercise assessed to one leg, pre and post test after 24 hours.	Passive stiffness increase after the eccentric exercised	18 males, half for stretching group and the rest for control group	Biodex dynamometer
(Ross and Engsberg, 2002)	Relationship of spasticity and strength at the ankles compared with the knees in those with cerebral palsy	Pearson correlation and t-test	No relation between spasticity and strength either within the same or opposing muscle groups at the knee and ankle joints in persons with cerebral palsy.	60 patients with spastic diplegic cerebral palsy (26 males, 34 females)	KinCom dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Singer et al., 2008)	Passive stiffness of cyclic stretching for hemiparetic subjects.	Comparisons were made between hemiparetic and normal subjects	No significant difference between groups. Plantarflexor resistive torque was reduced in all limbs following cyclic stretching regardless of stretch velocity	17 hemiparesis patients and 10 healthy subjects	Computer controlled, not specified
(Sobolewski et al., 2013)	Influence of maximum range of motion and passive stiffness on the viscoelastic stretch response	Use dynmomemeter and ultrasonic probe	Passive stiffness but not maximum range of motion influences the acute viscoelastic response to passive stretching	37 healthy men	Biodex dynamometer
(Sobolewski et al., 2014)	Age on the viscoelastic stretch response of ankle	Group comparison of the stiffness of ankle joint	Passive stiffness was greater in the older men	22 men (mean age: 24) and 14 older men (mean age: 67)	Biodex dynamometer

Author (year)	Parameters of interest	Method	Findings	Subject	Equipment
(Tilp et al., 2012)	Tibialis anterior aponeurosis during passive movements and active isometric, concentric, and eccentric contractions in vivo	Use dynmomemeter and ultrasonic probe	During isometric contractions, aponeurosis lengths increased and decreased with increasing and decreasing forces during passive movements, aponeurosis lengths did not change significantly	9 healthy subjects (2 women and 7 men)	Biodex dynamometer
(Whitting et al., 2013)	Passive weighted and non-weighted bearing dorsiflexion, range of motion and stiffness	By Pearson correlation	Passive dorsiflexion stiffness was not strongly associated with dorsiflexion range of motion	48 physically active males	KinCom dynamometer

2.2.4 Review of MPJ active and passive stiffness dynamometers

There are three kinds of MPJ dynamometers in literature. The two dynamometers invented by Senda et al., (1999) and by Goldmann et al., (2013) are shown in Figure 2-11 and Figure 2-12 respectively. They were both designed to determine the maximum forces produced by the five toes in active plantar flexion and abduction. The third dynamometer (Rao et al., 2011) measured passive stiffness of first metatarsophalangeal joint. Known dorsiflexion torque was applied on a wrench to rise only the big toe, and the angular displacement and torque resistance was recorded by potentiometer and torque transducer respectively (Figure 2-13). However, this device could not provide precise rotation speed and repeatable motion due to its manually controlled mechanism. Besides, there was no control of the alignment of the foot in the device.

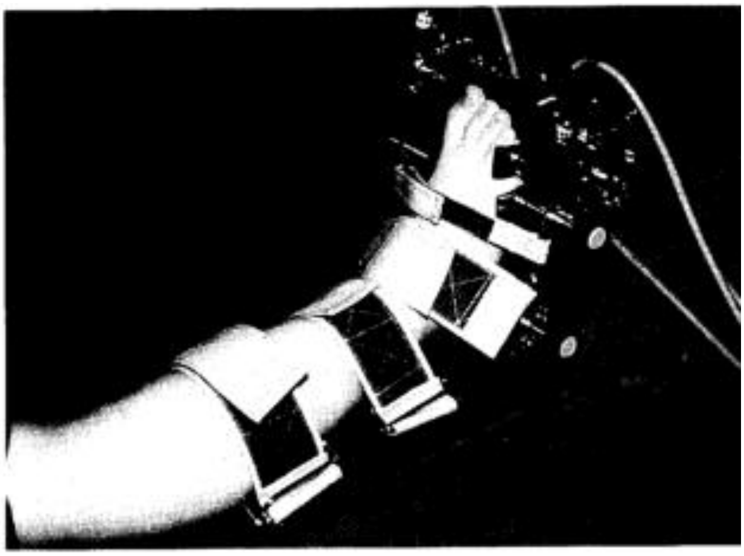


Figure 2-11 Toe dynamometer to measure toe flexion and abduction force (Senda et al., 1999)

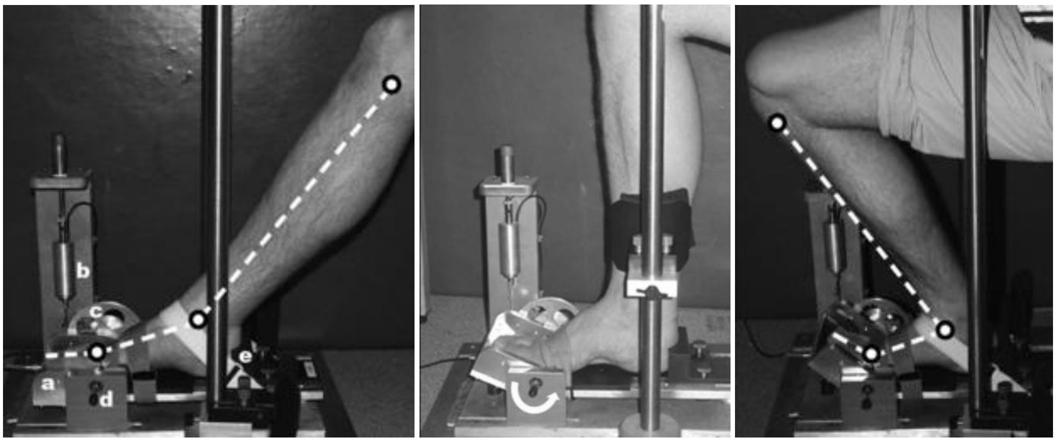


Figure 2-12 Toe dynamometer to measure toe flexion power in different posture (Goldmann et al., 2013)

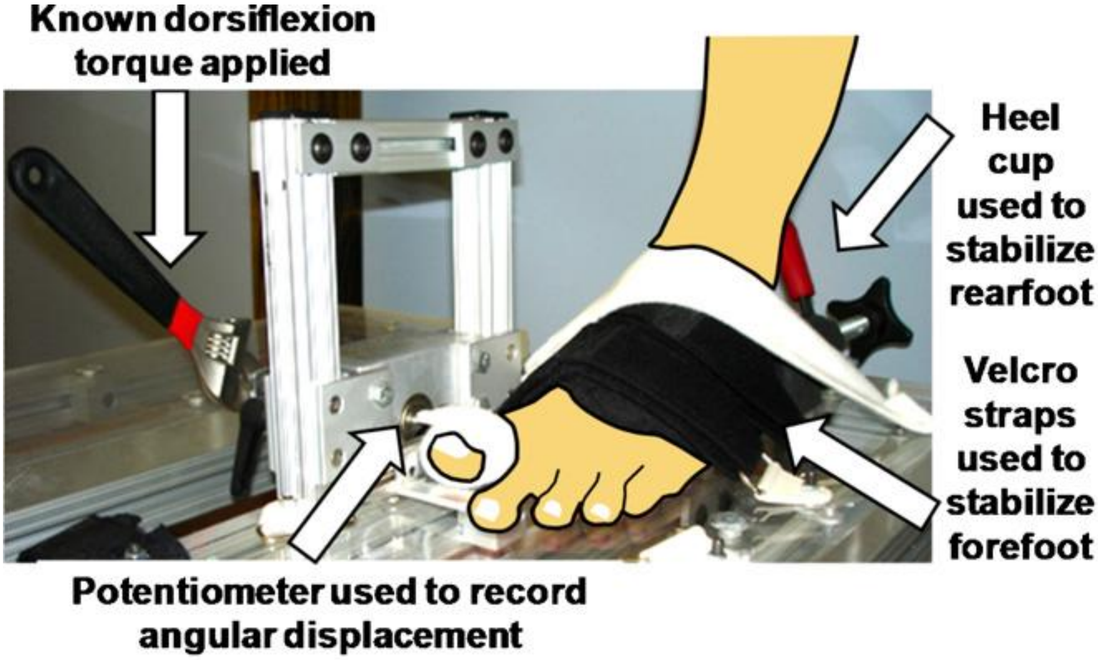


Figure 2-13 Custom made dynamometer for passive stiffness measurement of the big toe (Rao et al., 2011)

2.2.5 Summary of ankle joint and MPJ dynamometers

Large computer-controlled dynamometers were utilized to measure ankle joint passive stiffness in order to have repeatable passive ankle joint stiffness measurement. However, the large size of dynamometers limited clinical application. Kobayashi et al., (2011) invented a smaller dynamometer which was manually driven. However, constant passive motion of the joint was hardly achieved. Roa et al., (2011) only measured passive stiffness of the first metatarsophalangeal joint. Since the MPJ of the five toes are connected together by the transverse head of adductor hallucis and deep transverse metatarsal ligaments (Figure 2-4), it is more appropriate to measure the passive joint stiffness of all five metatarsophalangeal joints at the same time.

2.3 Spring-mass model of the leg

When human being is walking or running, the centre of mass (CoM) of the body is vertically oscillating and at the same time, the two legs are cyclically and alternatively compressing and extending (McMahon et al., 1987). The motion of both the CoM and the leg in the stance phase is often described as a “spring-mass model” (SMM). Damping was found negligible in human hopping (Rapoport et al., 2003). The model is thought as an ideal spring which is not influenced by weight, mass, or damping losses. It moves in one direction only (elongation or compression) and obeys the Hooke's law:

$$F=kd$$

where F, k and d represent the applied force, the spring stiffness and the displacement of compression or elongation of the spring respectively. The stiffness (k) is a constant factor. The displacement (d) is linearly proportional to the force (F). In the view of energy conservation, the leg compresses in length and stores energy in elastic soft tissue in the first half of stance, and then recoils in the second half of stance in order to push the body upwards and forwards (Cavagna et al., 1964, Delecluse et al., 1995, Dalleau et al., 1998, Farley et al., 1991, Farley and Gonzalez, 1996, Ferris et al., 1998, He et al., 1991, McMahon et al., 1987, McMahon and Cheng, 1990). The compression and extension motion of the leg is called leg spring. In addition, SMM is also applied to describe the vertical motion of the body's CoM in running and hopping, and is called vertical stiffness (K_{vert}). It was hypothesized that K_{vert} was proportional to running economy. The smaller the amplitude of vertical CoM oscillation is, the more efficient of walking or running would be (Inman and Eberhart, 1953). It was reported that both the leg stiffness (K_{leg}) and the K_{vert} were influenced by lower limb joint stiffness (Farley and Morgenroth, 1999, Hobara et al., 2009). Since this mechanical stiffness of human legs can be a key influence on performance, it has been an area of interest for researchers for many years.

2.3.1 Vertical stiffness

K_{vert} represents the vertical motion of the CoM during hopping (vertical jumping) or running. It is defined as the ratio of the maximum vertical reaction force (F_z) of the ground to the CoM displacement. It occurs in the middle of landing phase for both hopping and running. Cavagna (1975) demonstrated how displacement of CoM could be determined with force platform only. When a subject is hopping on a force platform, vertical reaction force (F_z) exerted by the feet can be expressed as

$$F_z = BW + ma_z$$

where BW, m and a_z represent the body weight of subject, body mass and acceleration of CoM in the vertical direction respectively. Hence, displacement of CoM (Δy) can be determined by the double integration of a_z:

$$a_z = \frac{BW - F_z}{m}$$

The acceleration curve is integrated over time to get the velocity (v'), i.e.

$$v' = \int a_z \, dt + C$$

Since a human subject keeps constant jumping height during the hopping testing, the constant C is assumed to be zero. The hopping velocity (v') in upward and downward movement are equal over time. The velocity for continuous hopping is equal to v' minus averaged v', i.e.

$$v = v' - \overline{v'}$$

The displacement of vertical CoM (Δyx) is determined by the integration of v over time i.e.

$$\Delta y = \int v \, dt + C$$

Again, a subject keeps constant jumping height during the hopping testing, the constant C is assumed to be zero. Accordingly, McMahon and Cheng (1990) proposed the following common method to determine the k_{vert} during hopping:

$$K_{vert} = \frac{F_{z(max)}}{\Delta y_{(max)}}$$

where $F_{Z(max)}$ and $\Delta y_{(max)}$ stand for the maximum vertical force and the maximum displacement of the CoM. In Figure 2-14 for example, the K_{vert} of the spring was equal to F_{peak}/ y_c.

McMahon et al. (1987) proposed that CoM is oscillating on an ideal spring in simple harmonic motion, and K_{vert} can be expressed as

$$K_{vert} = m\omega^2$$

where ω is the natural frequency of the oscillation of the CoM and m is the body mass. In McMahon's method, reaction force is first collected by force platform when running. Then, the vertical velocity is determined by single integration of the vertical acceleration ($a_z = (BW-F_z)/m$). Together with the initial contact time, the natural frequency of oscillation ω can be solved by the equation:

$$\frac{F}{mg} = \left(\frac{v\omega}{g}\right)\sin\omega + 1 - \cos\omega t$$

where v represents the vertical velocity of the CoM.

An alternative method to determine ω was suggested by Cavagna et al. (1988). The duration of the vertical force over the body weight in the contact phase is defined as the effective contact phase (t_{ce}). Since the point mass in SMM is oscillating in sine-wave pattern, the loading period is equal to the unloading period. As a result, the duration of t_{ce} is equal to half of the whole oscillation period (P), i.e. t_{ce}=P/2. The ω in SMM can be alternatively expressed as:

$$\omega = 2\pi f \text{ or } \omega = 2\pi(1/P)$$

where f is the oscillating frequency which is equal to the reciprocal of P. Then

$$K_{vert} = m(2\pi/P)^2 \text{ or } K_{vert} = m(\pi/t_{ce})^2$$

Figure 2-15 shows the half period of oscillation in force-time curve in stance phase.

In summary, both methods presented by McMahon et al. (1987) and Cavagna et al. (1988) determine the K_{vert} with force platform and confirm the formula $K_{vert} = m\omega^2$, although the ω of each method is calculated in different approaches.

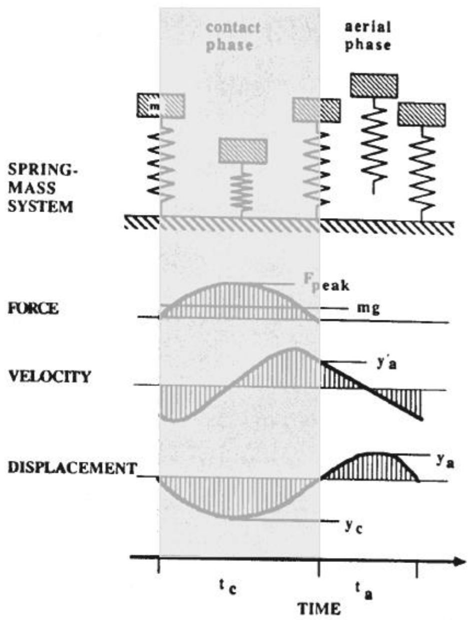


Figure 2-14 Force, velocity and displacement act against time of point mass in SMS during hopping (Blickhan, 1989)

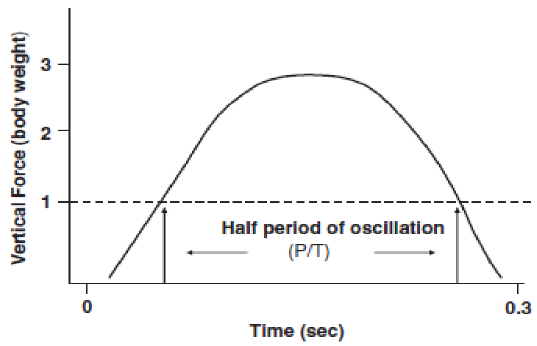


Figure 2-15 Force-time curve is related to body mass. The half-period of oscillation is measured when the force is above body weight during the standing phase of a bouncing gait (Brughelli and Cronin, 2008)

However, all the above methods require the use of force platform which is expensive and inconvenient for field test. Dalleau et al. (2004) invented an alternative method to calculate the K_{vert} in vertical hopping without using a force platform The vertical force is modelling as a sine wave since the force of SMM is oscillating in simple harmonic motion. K_{vert} is expressed as:

$$K_{vert} = \left(\frac{m\pi(t_a + t_c)}{{t_a}^2 \left(\frac{t_a - t_c}{\pi} - \frac{t_c}{4} \right)} \right)$$

where t_a, t_c and m represent aerial time, contact time and body mass. To experimentally validate the new method, the researchers put a tiny pressure sensor mate on a force platform to collect the F_z ,t_a and t_c simultaneously during hopping. Then t_a and t_c were collected by the pressure sensor and substituted to the proposed formula to yield K_{vert}. The F_z collected by the force platform was used to calculate the $\Delta y_{(max)}$ and hence K_{vert} by the conventional double integration method proposed by Cavagna et al. (1975) and McMahon and Cheng (1990) as reference. The researchers found that the K_{vert} determined by the two methods were highly correlated (r=0.91, p<0.001) and considered this new method was valid to get K_{vert} with only t_a and t_c.

Morin et al. (2005) also developed and validated a formula without using force platform data to determine K_{vert} from body mass, t_a and t_c, rather than force platform. Similar to previous methods, the force-time curve was assumed as a sine function: $K_{leg} = F_{z(max)}/\Delta y_{(max)}$ where

$$F_{z(max)} = mg\frac{\pi}{2}\left(\frac{t_c}{t_a}+1\right)$$

$$\Delta y_{(max)} = \frac{F_{z(max)}t_c^2}{m\pi^2} + g\frac{t_c^2}{8}$$

In the experiment, 2 pieces of pressure switches were put under an insole and located to the heel and the ball of foot (underneath 1^{st} to 5^{th} MPJ) to measure t_a and t_c. In order to obtain reference results for comparison, pressure sensors were mounted on both treadmill and ground to capture vertical reaction force. In both cases, conventional force platform method proposed by Cavagna (1975) and McMahon and Cheng (1990) and the new sine-wave method were used to calculate results for comparison. The reference-model error for treadmill and over-ground running were 0.12% and 2.30% respectively. The method was more economical and reliable comparing to the conventional method for K_{vert} measurement.

2.3.2 Leg stiffness

Leg stiffness (K_{leg}) refers to the ratio between maximum change in leg length (ΔL) and maximum ground reaction force $F_{z(max)}$, i.e.

$$K_{leg} = F_{z(max)}/\Delta L$$

Using a force platform, McMahon and Cheng (1990) provided a solution for K_{leg} measurement. Figure 2-16 illustrates the model of leg stiffness. The total leg compression is expressed as the following:

$$\Delta L = \Delta y + L_0 (1 - \cos \theta_0)$$

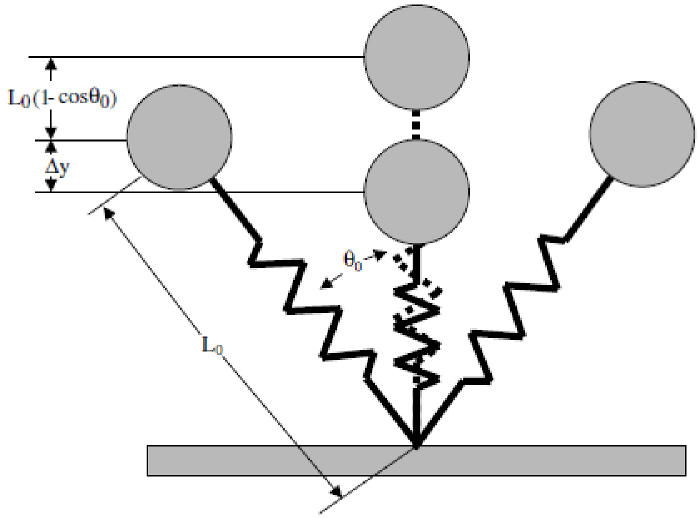


Figure 2-16 Leg stiffness model used to determine K_{leg} when leg contacted with the ground during running (Butler et al., 2003)

where θ_0 is the angle of leg from heel strike to mid-stance and L_0 is the leg length in standing position, i.e. distance from ground to greater trochanter. Similar to K_{vert} calculation, $\Delta y_{(max)}$ can be obtained by double integration of net acceleration of the CoM, a_z=((F_z - BW)/m). In terms of trigonometry,

$$\theta_0 = \sin^{-1}\left(\frac{vt_c}{L_0}\right)$$

where v is the forward velocity and t_c is the contact time duration. v can be determined by the given sampling frequency of the force platform and the contact time t_c. Consequently, K_{leg} can be determined with force platform.

A force platform is expensive and complicated to set up. Moreover, a single and standard force platform can only capture K_{leg} of a single step in forward running. It is very difficult for most laboratories to set up a series of force platforms in a row to capture ground reaction force of successive steps and check the change of K_{leg} in sequence. To solve this problem, Morin et al. (2005) determined K_{leg} by adopting affordable foot switch and radar gun to measure t_a, t_c and v continuously. Similar to the proposed method for K_{vert} measurement, the F_z was modelled as a sine wave during forward running, and K_{leg} was determined with the following equations:

$$K_{leg} = \frac{F_{z(max)}}{\Delta L} \qquad F_{z(max)} = mg\frac{\pi}{2}\left(\frac{t_c}{t_a}+1\right)$$

$$\Delta y_{(max)} = \frac{F_{z(max)}t_c^2}{m\pi^2} + g\frac{t_c^2}{8} \qquad \Delta L = L - \sqrt{L^2 - \left(\frac{vt_c}{2}\right)^2} + \Delta y_{(max)}$$

where v is the forward velocity recorded by radar gun. Similar to the K_{vert} measurement, t_c and t_a were recorded by 2 pressure switches positioned under the heel and the ball of foot (region underneath 1st to 5th MPJ) for both over-ground and treadmill running experiment. The researchers found that the rates acquired were 0.67% and 6.93% respectively and were lower than those determined by the double integration of the vertical acceleration over time proposed by Cavagna (1975) and McMahon and Cheng (1990).

The two methods aforementioned for K_{leg} measurement determine ΔL indirectly. Only three studies directly used force platform to record Fz and then measured ΔL with either 3D motion capture system (Arampatzis et al., 2006, Grimmer et al., 2008) or 2D camera (Rapoport et al., 2003).

Table 2-2 Equipment and formula for stiffness determination

	Equation	Equipment	Reference
Vertical Stiffness, K_{vert}	$K_{vert} = m\omega^2$	Force platform	McMahon et al., (1987)
	$K_{vert} = m(2\pi/P)^2$	Force platform	Cavagna et al., (1988)
	$K_{vert} = F_{Z(max)}/ \Delta y_{(max)}$	Force platform	McMahon and Cheng (1990)
	$K_{vert} = m((\pi(t_a+t_c))/(t_c^2(((tv+t_c)/\pi))-(t_c/4))))$	Foot switch	Dalleau et al., (2004)
	$K_{vert} = F_{Z(max)}/ \Delta y_{(max)}$ $F_{Z(max)} = mg(\pi/2)(t_c/t_a+1)$; $\Delta y_{(max)} = (F_{z(max)}t_c^2/m\pi^2)+g(t_c^2/8)$	Foot switch	Morin et al., (2005)
Leg Stiffness, K_{leg}	$K_{leg} = F_{Z(max)}/ \Delta L$ $\Delta L = \Delta y + L (1 - \cos \theta)$	Force platform	McMahon and Cheng (1990)
	$k_{leg} = FZ(max)/ \Delta L$ $F_{Z(max)} = mg(\pi/2)(t_c/t_a+1)$; $\Delta L=L-(L^2-(vt_c/2)^2)^{0.5}+\Delta y_{(max)}$; $\Delta y_{(max)} = (F_{z(max)}t_c^2/m\pi^2)+g(t_c^2/8)$	Foot switch	Morin et al., (2005)
	$K_{leg} = F_{Z(max)}/ \Delta L$ ΔL: distance of reflective marker on greater trochanter to the ground on mid-stance	Force platform, high-speed camera	(Arampatzis et al., 2006, Grimmer et al., 2008) (2003)

2.3.3 Relationship of different kinds of stiffness

A vast number of researchers investigated the relationship among different kinds of stiffness in running and jumping. Both intrinsic and environmental factors affect K_{leg} and K_{vert}. Dalleau et al., (2004) found that the K_{vert} increased from 21.8kNm^{-1} to 68.4kNm^{-1} with hopping frequency changing from 1.8Hz to 4Hz. When the hopping frequency reduced, the landing time increased and hence increased the time and displacement for vertical movement of center of mass. Finally, the vertical stiffness reduced. Conversely, when the frequency increased, the landing time decreased and hence decreased the time and displacement for vertical movement of center of mass. Finally, the vertical stiffness increased. He et al., (1991) investigated the K_{vert} and K_{leg} at different running speeds. Four subjects ran on a force plate-mounted treadmill at 5 velocities starting from 2.0 m/sec and then increasing to 6.0 m/sec. The K_{vert} increased with velocity from 20kN/m to 50kN/m but the K_{leg} remained unchanged at 11.3kN/m. Avogadro et al., (2004), Cavagna et al., (2005) and Morin et al., (2005) also reported that the K_{leg} remained constant while the K_{vert} increased along with running velocity increased from slow to moderate velocities. Farley et al. (1993) examined the K_{leg} and K_{vert} of animals with different body sizes in a range of velocities and found that animals of different body sizes had constant K_{leg} but the K_{vert} increased with increasing velocity and stiffness of the ground. Daniel et al., (1997) reported that the K_{leg} increased as high as 3.6 times to accommodate soft ground. Farley et al., (1999) also demonstrated that the K_{leg} approximately increased twice when the subjects hopped on relative soft ground.

K_{vert} and joint stiffness increases in high running speed (Arampatzis et al., 1999, Luhtanen and Komi, 1980, Günther and Blickhan, 2002). K_{leg} primarily depends on ankle joint stiffness in preferred hopping height at 2.2Hz (Farley and Morgenroth, 1999). However, knee joint stiffness is a major determinant of K_{leg} in maximal hopping (Hobara et al., 2009). Some other studies found that ankle joint stiffness remained constant in running, but knee joint stiffness increased with rising running speed (Arampatzis et al., 1999, Günther and Blickhan, 2002, Kuitunen et al., 2002, Stefanyshyn and Nigg, 1998). Stefanyshyn and Nigg (1998), Chelly and Denis (2001) and Butler et al.(2003) suggested that running speed might be enhanced with greater K_{vert} or knee joint stiffness.

2.3.4 Relationship between foot morphology and vertical stiffness

The arches of the foot are partially sustained by toe flexors such as abductor halluci, abductor digiti minimi, flexor digitorum brevis and flexor digitorum longus. It is reasonable to assume that the strength of toe flexor muscles is related to the arches of the foot. Rao and Joseph (1992) conducted a survey on 2300 children and revealed that the possibility for children who frequently use footwear to be flatfoot was 8.6%, while for those who usually walked barefoot was 2.8% ($p < 0.001$). It was believed that the unshod condition enhanced the development of foot muscles for arch support and hence the possibility of flatfoot in barefoot or minimum shoes (such as sandal) condition were significantly lower compared to shod condition. In addition to muscle strength, the mechanical properties of toe flexors were related to the arch of foot. The stiffness of the 1st MPJ was evaluated by a custom-made device

(Rao et al., 2011). The researchers recruited 61 individuals and classified their feet structure into three categories as high, normal and low arches according to quantitative measures such as malleolar valgus index and arch height index. They successfully demonstrated that the 1st metatarsophalangeal joint late flexibility in high degree of dorsiflexion during sitting position was significantly higher in individuals with low arch compared to high arch structure ($p=0.03$), and that the 1st metatarsophalangeal joint flexibility in high degree of dorsiflexion during standing was also significantly higher in individuals with low arch compared to normal arch structure ($p=0.03$). To correlate foot morphology with leg stiffness, Williams et al. (2003) compared the K_{leg} of 20 high-arched runners with 20 low-arched runners and found that the high-arched runners exhibited greater K_{leg} than the low-arch runners when running. High-arch foot is usually associated with decreased pronation in stance phase, and hence decreases knee flexion excursion. The more the knee extended, the stiffer the leg would be.

2.3.5 Stiffness and locomotion efficiency

Although the stiffness of the lower extremities (K_{leg}, K_{vert}, active and passive joint stiffness) and their relationships of the locomotion performance have been studied for decades, the results are controversial. Many previous findings support that the lower limb stiffness increases with demands of activities. Some studies reported that K_{vert} increased with higher hopping frequency (Farley et al., 1993, Granata et al., 2002) and hopping height (Arampatzis et al., 1999). K_{vert} also increases with running speed (Arampatzis et al., 1999, Seyfarth et al., 2002, Stefanyshyn and Nigg, 1998). Chelly and

Denis (2001) and Butler et al., (2003) suggested that running speed might increase with greater K_{vert} or knee joint stiffness. Regarding to running economy (energy saving), it was reported that lower limb stiffness improved running economy (Kerdok et al., 2002, Dutto and Smith, 2002, McMahon and Cheng, 1990).

Taylor & Beneke (2012) examined the K_{leg} and K_{vert} of 3 fastest runners in 100m dash and found that both stiffness of the champion, Usain Bolt, were significantly lower than his competitors. The researchers believed that the high compliance (inverse of stiffness) of Usain Bolt facilitated the storage and utilization of elastic energy during the stretch shortening cycle. Seyfarth et al. (2000) suggested that there was an optimal mechanical stiffness for long jumping and the improvement of this stiffness would not enhance jump distance.

2.3.6 Stiffness and injury

Although the degree of limb stiffness might enhance performance, it would also increase the risk of injuries. Since high leg stiffness increases peak force or decreases lower limb flexion, it would also increase the possibility of lower extremity shock (Lafortune et al., 1995). The shock, high peak force and high loading rates are also related to bone injuries including stress fracture and osteoarthritis (Grimston et al., 2010, Radin et al., 1978, Burr et al., 1985). On the contrary, high compliance (low stiffness) would cause soft-tissue injury because it might allow excessive joint motion. It was reported that runners with low K_{leg} had more soft tissue injuries than high leg stiffness counterparts

(Williams et al., 2003). Granata et al., (2002) demonstrated that the degree of knee stiffness in female was less than male in hopping and this might explain why female had higher percentage of knee ligamentous injuries.

2.4 Physiology of sub-maximal running

Unlike sprinting, sub-maximal running refers to a person running for a relative long time without fatigue, or running aerobically. The speed for sub-maximal running is specific for individual, depending on fitness level and age etc. of the subject. It is necessary to monitor individual subjects running in aerobic state by fatigue index in sub-maximal running exercise. The common fatigue index includes target heart rate (Bunc et al., 1988) and respiratory exchange ratio (RER) (Astorino et al., 2000).

2.4.1Target heart rate

Target heart rate is the maximum number of heart beat per minute (bpm) for exercise intensity under aerobic threshold. The range mainly depends on age. However other criteria such as sex and individual physical condition are also included. The formula is 220 minus age. The intensity can be calculated as a range of percentage. For example, for a person aged 26, the target heart rate would be

$$220-26=194 \text{ bpm}$$

For 90% intensity of target heart rate, it would be

$$(220-26) \times 0.9=174\text{bpm}$$

2.4.2 Respiratory exchange ratio

The ratio of oxygen breathe in to carbon dioxide breathe out is called the respiratory exchange ratio (RER). This indicates whether carbohydrate or fat is being metabolized to supply energy to the body. In resting stage, the average RER which indicates that fat is the predominant fuel source is about 0.70. During exercise, a RER of 0.85 suggests a mix of fat and carbohydrates. As the intensity of exercise becomes higher, RER can be 1.00 or above and such a condition indicates that carbohydrate is the predominant fuel source. In very intense exercise, human body needs more oxygen to oxidize the blood lactic acid, and the level of bicarbonate changes. These adjustments will buffer additional non-respiratory carbon dioxide and will cause higher RER. A RER greater than 1.15 is often used as a secondary endpoint criterion for a maximal oxygen consumption test (Astorino et al., 2000).

Oxidation of a molecule of fatty acid:

$$23\ O_2 + C_{16}H_{32}O_2 + 129ADP + 129P_i \Rightarrow 16\ CO_2 + 16\ H_2O + 129\ ATP$$

$$RER = VCO2/VO2 = 16\ CO2/23\ O2 = 0.7$$

Oxidation of a molecule of carbohydrate:

$$6\ O_2 + C_6H_{12}O_6 + 38ADP + 38P_i \Rightarrow 6\ CO_2 + 6\ H_2O + 38\ ATP$$

$$RER = VCO_2/VO_2 = 6\ CO_2/6\ O_2 = 1.0$$

The above equations indicate that oxidation of fatty acid requires less oxygen than oxidation of carbohydrate, but convert more adenosine diphosphate (ADP) to adenosine triphosphate (ATP).

2.4.3 Equipment of running economy measurement

RE measures the energy cost while running aerobically. It is usually indicated by oxygen consumption normalized to BW and running distance (or time), and expressed as ml•kg^{-1}•m^{-1}, while some researches expressed as ml•kg$^{-0.75}$ •min^{-1}, since oxygen consumption may be better related to the -0.75 power of body mass rather than reciprocal of body mass(Bergh et al., 1991). Direct and indirect calorimetries are used to measure RE. For direct calorimetery, the subject is enclosed with a small sealed chamber and the amount of heat is measured. Since the small chamber is heat isolated and the equipment is sensitive enough to detect the heat change, the system would be expensive. For indirect calorimetry, the amount of heat produced by the subject is calculated by determination of the amount of oxygen consumed and the amount of carbon dioxide released. Indirect calorimetery can be either closed circuit or open circuit. Closed circuit indirect calorimetery is similar to direct calorimetery, and in such a case, the subject is enclosed within a small chamber, and intake oxygen, expired oxygen and carbon dioxide are all measured to estimate RE. In open circuit calorimetery, the subject put on a mask to collect the volume of air, oxygen and carbon dioxide to determine RE. Expensive open circuit calorimetery can be small and light in weight so that the system is worn on the back or chest with wireless data emitter. Subjects can wear the gas analyzer and run on treadmill or run around for field test (Figure 2-17).

Figure 2-17 Open circuit calorimetery, spiroergometer METAMAX 3B R2
(Source: http://www.vo2max.pl/page,ergospirometry mobilne)

2.5 Summary of Literature review

The MPJ play a major role in increasing the weight-bearing area (i.e., reduced plantar pressure) and enhancing body balance in walking and running (Hughes et al., 1990, Tanaka et al., 1996, Mueller et al., 1997). In sports performance research, some studies reported that MPJ flexor strength training would improve jumping (Goldmann et al., 2013) and running performance (Hashimoto and Sakuraba, 2014). However, research of the MPJ were very limited compared to other lower limb joints such as the ankle. Up to date, computerized dynamometer can only be applied to the hip, knee and ankle but not MPJ. Few dynamometers for MPJ torque measurement were controlled manually. Computerized dynamometer for MPJ torque measurement would provide meaningful insight for biomechanics researches and clinical applications. For K_{leg}, K_{vert}, and RE relationships, it was considered that high

K_{leg} indicated more capacity of elastic energy storage in muscle-tendon unit of the leg (Dalleau et al., 1998) and high Kvert reduced the energy expense for vertical motion of CoM (Heise and Martin, 2001). However, the effect of MPJ stiffness on RE is not found in literature. It has been hypothesized that stronger toe muscles would be possible to reduce contact time (tc) by enhancing balance. Since both K_{vert} and K_{leg} are beneficial to RE and inversely proportional to tc, MPJ enhancement might improve K_{vert}, K_{leg} and eventually RE.

CHAPTER 3. Development of Ankle Foot Dynamometer

3.1 Introduction

The plantar flexor muscles of human toes play important roles in locomotion. In walking and running, the forefoot generates force by pushing against the ground to provide propulsion during late stance (Mann and Hagy, 1979). In athletic performance, increase of toe flexor strength leads to increase of horizontal jumping distance in adults (Goldmann et al., 2013). The relative toe flexor strength (toe flexor strength divided by body weight) is positively correlated with sprinting, standing broad jump, repeated side-step, and rebound jump performance in children (Morita et al., 2014). Weakness of intrinsic foot muscles is a likely cause of toe deformities, such as hammer toes and claw toes (Garth and Miller, 1989, Caselli and George, 2003, Myerson and Shereff, 1989, Van Schie et al., 2004). The frequency of falls in the elderly also increases with toe weakness and deformity (Mickle et al., 2009), therefore it is of genuine importance to reliably determine MPJ stiffness, which reflects the strength of toe flexor muscles. This, should help to assess the effectiveness of toe muscle training regimens, or to evaluate the benefit of surgical and conservative interventions aiming to correct toe deformities.

Ankle joint stiffness also plays an important role during locomotion. Regarding performance aspects, active stiffness of the ankle joint decreased by 32.7% after a seven-week power training, but passive stiffness increased by 58.2% (Cornu et al., 1997). Active ankle joint stiffness reduction seems desirable for the rate of force development, whereas increase in passive stiffness should be advantageous for reuse of elastic energy stored in the muscular structures. It is suggested that eccentric exercise causes an increase in passive stiffness, in turn improving athletic performance during running and jumping (Lindstedt et al., 2002, LaStayo et al., 2003, Reich et al., 2000). Therefore, ankle joint stiffness needs to be considered as a parameter to evaluate leg muscle training regimens, and to serve as a predictor of physical performance. In pathological scenarios, excessive or contractual ankle joint stiffness resembles a general complication of common clinical disorders, such as spasticity or prolonged immobilization after stroke or plaster casting (Kwah et al., 2012, Vattanasilp et al., 2000, Nightingale et al., 2007, Taylor and Allum, 1988). Consequently, it will reduce the range of motion of ankle joint or even cause drop foot symptom and hence are responsible for locomotion insufficiencies (Taylor and Allum, 1988). Ankle contractures are treated by several methods, including active muscle strength training (Selles et al., 2005) and passive stretching exercise (Prabhu et al., 2013). Thereby, ankle joint stiffness should be monitored and used to evaluate the effectiveness of the applied method for individual patient (Zhang et al., 2002, Moseley and Adams, 1991).

Dynamometers, offered by Biodex and Cybex, are used to quantitatively determine strength and stiffness of ankle (Arampatzis et al., 2005, Akhbari et al., 2007) , knee (Peixoto et al., 2011, Nordez et al., 2009), elbow(Nesterenko et al., 2010) and shoulder (Edouard et al., 2013) joints and their development due to sport and rehabilitation programs during decades. A manually operated dynamometer device was used to measure the common strength of maximum isometric contraction of all toes (Goldmann et al., 2013). Recently, the passive stiffness of the big toe of a low arched foot during dorsi-flexion was found to have a significantly higher late flexibility of the 1st MPJ compared to those with high arch foot in sitting position, and the big toe flexibility of low arch foot was significantly higher than that in normal arch foot in standing positions (Rao et al., 2011). However, the angular speed and displacement of the big toe flexion might have varied under manual control of which reliability assessment was not reported. Moreover, all current dynamometers did not provide valid, repeatable methods to align their rotating axes to the anatomical axis of the joint(s). Since the passive torque of MPJ is considered small compared to that of big joints such as ankle or knee, potential measurement errors due to misalignment of the foot may be relatively large and hence deteriorate validity and reliability of measurements.

In an effort to improve the current standards of MPJ and ankle torque measurements, the aim of this research was to develop a dynamometer which provides objective, quantitative torque measurements for MPJ and ankle joint. Specifically, this involved an accurate alignment method regarding the device's rotating axis relative to the corresponding anatomical joint axis and computer driven and controlled angular speed and displacement characteristics. Furthermore, the reliability of the device was examined by within-day and between-day test-retest experiments.

3.2 Overview of the dynamometer

The custom made dynamometer consists of a console and a detachable foot attachment device. The console consists of a swing cradle, driven by a motor-sensor unit, which includes a computer-controlled servo motor (Cool muscle Inc, Model: RCM1-C-23L20-C-RT3, Canada), gearbox (Cool muscle Inc, Model: RGP60-80-NEMA23, Ratio 80:1 and 3 Stage, Canada), a torque limiter (R+W Inc, Model: SK2/ 15/75(W), USA), a calibrated torque transducer (Mountz Inc, Type RTSX 200 IA II, Range 2.3-22.6Nm and sampling frequency 100Hz, Germany) and a pair of vertical laser line projectors. The selected combination of motor and step-down gearbox allows a maximum speed of 45°/s and a maximum output torque generation of 65Nm. The torque limiter connecting the gearbox and the torque transducer will reversibly disengage the power transmission when the torque of cradle exceeded the preset limit of 20Nm to avoid danger. Another end of shaft of torque transducer connects the cradle, which is supported by a pair of rotary bearing on the rigid frame. The swing speed and the angular displacement of the cradle are driven by the computer-controlled motor, and the resistances of the rotation, or torque, of the cradle is measured by the torque transducer. The computer-controlled motor and the torque transducer were controlled by two independent software and there were not any feed-back control system in the dynamometer. The laser lines vertically pass through the axis of shaft of motor-sensor unit in order to visualize the rotating axis of cradle for device and MPJ/ankle joint alignment (Figure 3-1).

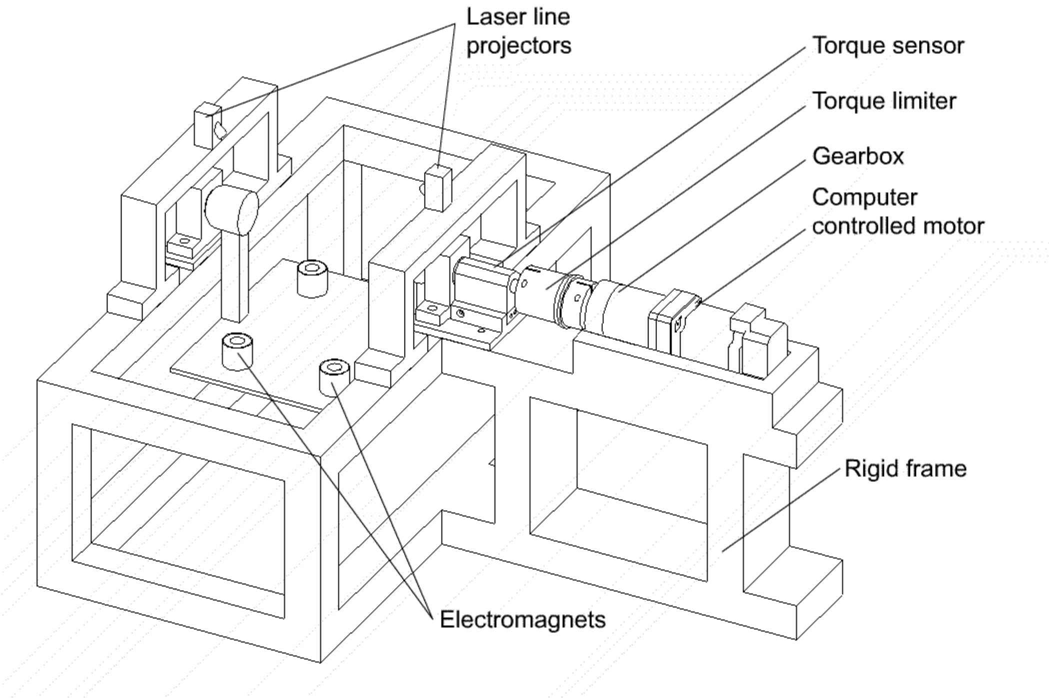


Figure 3-1 The console of ankle-foot dynamometer

The foot is fixed on a detachable foot platform which is then placed on the cradle (Figure 3-2). Electromagnets of the cradle can attach the magnetized metal plate on the base of foot platform and hence enabled the footplate to be instantaneously fixed to or released from the console.

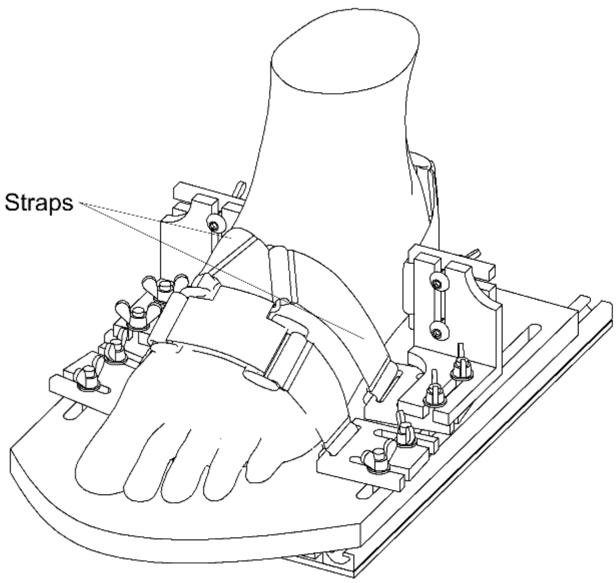


Figure 3-2 The foot was fixed on a detachable foot platform by Verlco-strap

The vertical foot position can be adjusted by changing the thickness of footplate between the top plate and bottom magnetized plate (Figure 3-3).

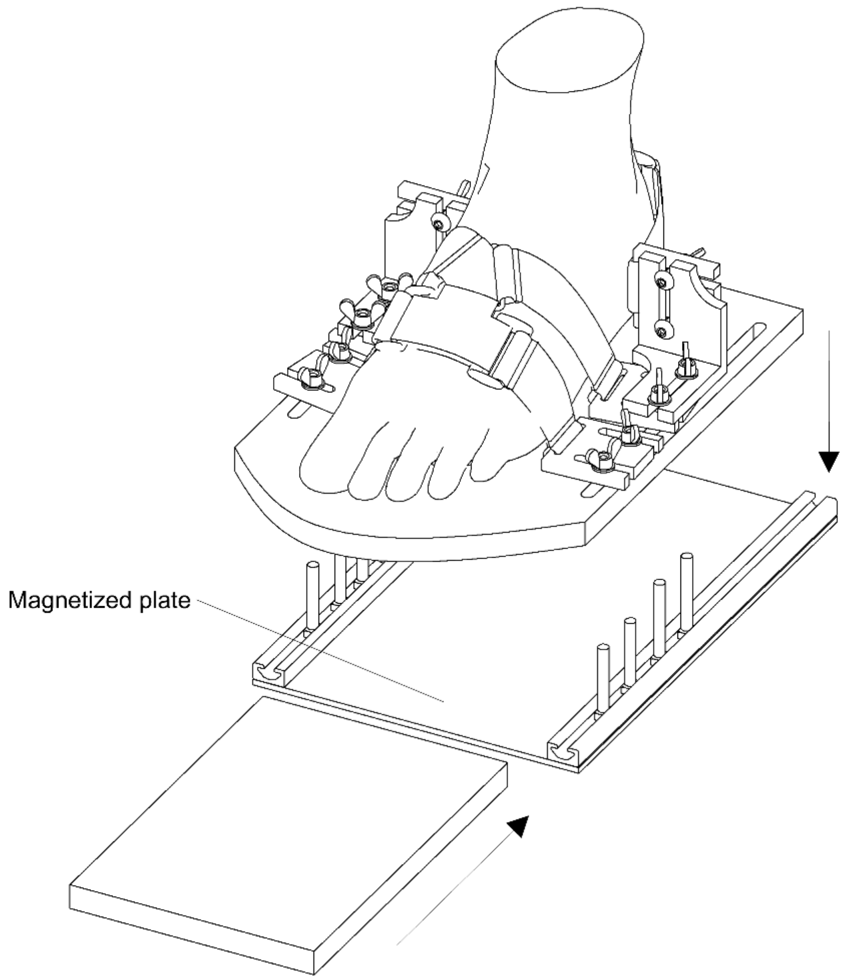


Figure 3-3 The height of foot platform is adjustable by inserting different thicknesses of rigid plate in the middle. The bottom plate is an magnetized metal plate for attachment of electromagnets on the cradle.

Thus, MPJ or ankle joint axis matches the height of the rotating axis of cradle independent of foot anatomy (Figure 3-4). The dynamometer supports measurements of varied foot sizes ranging from 6 to 14 in US scale.

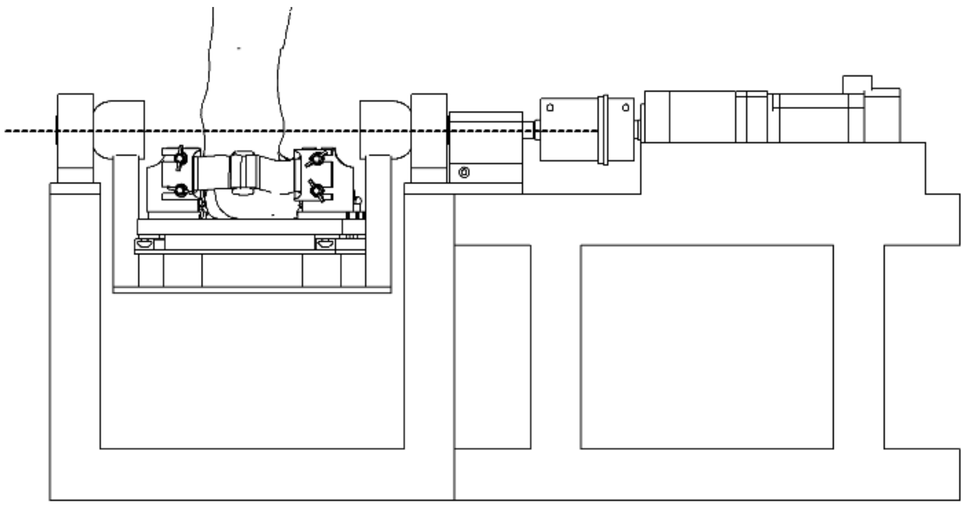


Figure 3-4 Suitable thickness of rigid plates were inserted to the middle of footplate so that ankle joint axis is aligned to the rotating axis of the dynamometer

For MPJ torque measurement, an additional rigid plate with electromagnets was fixed on the console to support the foot platform (Figure3-5).

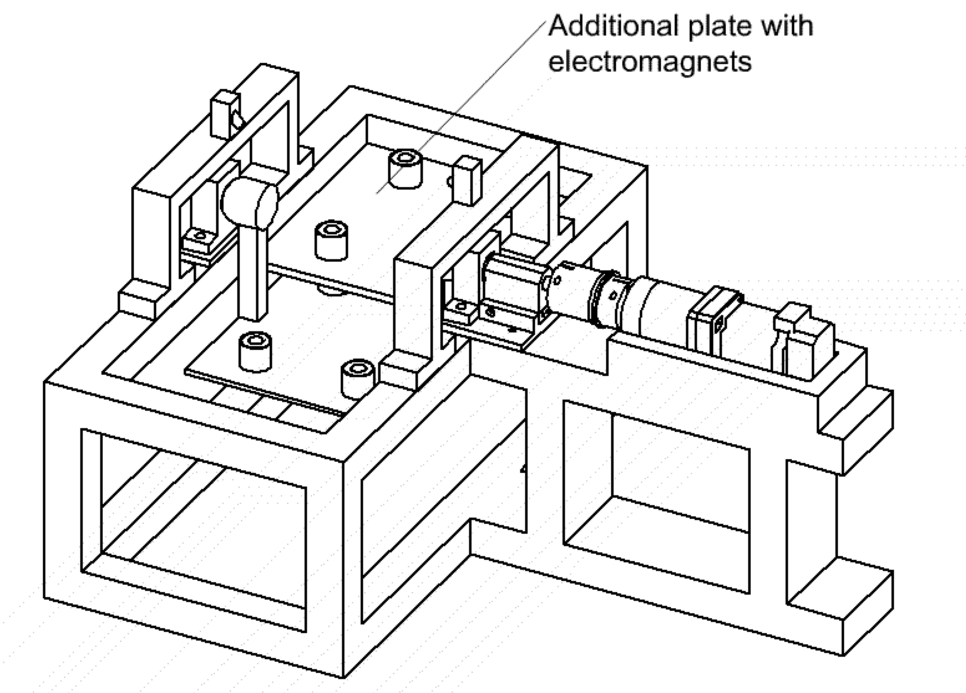


Figure 3-5 In MPJ configuration, additional plate with electromagnets was installed on the console to support the static foot platform.

In addition, a toe platform, structurally similar to the foot platform was placed on the cradle to support the toes (Figure 3-6). A flexible and soft plastic film was put on top of the cradle and foot platform in order to eliminate the friction between the toes and cradle during swing. It was important because the irritating friction would elicit active movement of the toes. In addition, the film provided support and preventing the toes from falling the gap between cradle and the foot platform. The contour of top plate of toe platform matched the front curved margin of foot platform in order to combine the footplates and provide a flat surface to support the whole foot.

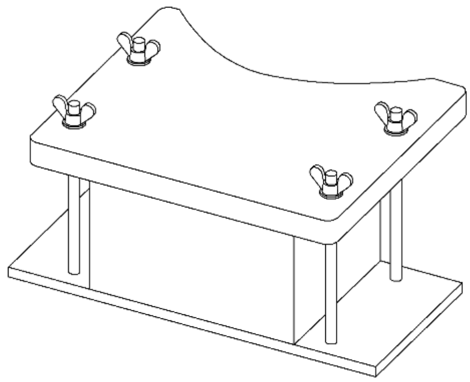


Figure 3-6 In MPJ configuration, toe platform was used to support the toes and attached to the swing cradle. This structure was similar to foot platform, with removable block in the middle to control the height. Magnetized metal plate was used for attachment of electromagnets on the cradle.

The MPJ axis was defined as a horizontal straight line passing through the medial and lateral apices of first and fifth metatarsal heads. The foot first placed on the foot platform with matched front margin, so that the whole foot excepting all the toes could step on it (Figure3-7).

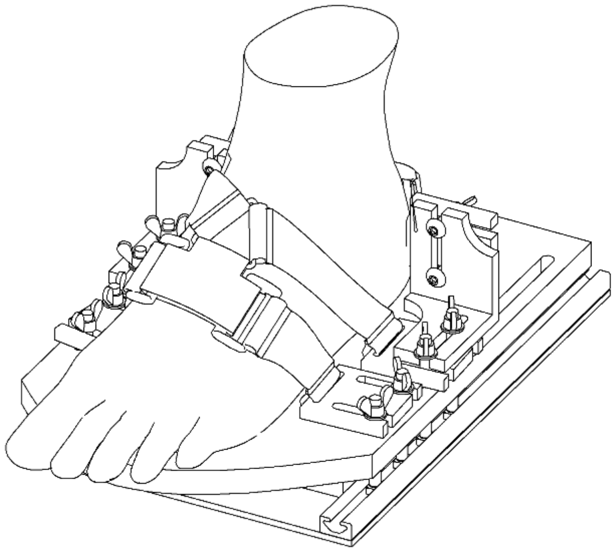


Figure 3-7 In MPJ configuration, the foot was fixed on the foot platform excepting that the toes were stepping out of the front edge

After the height of foot platform and toe platform was adjusted so that the top surface of both platforms were of the same level and the MPJ axis reached the altitude of rotating axis of the cradle by aforementioned plate inserting method. The foot was fasted on the foot platform by Velcro straps. When the electromagnets were turned off, the foot and toe platforms were slided on the electromagnets until the vertical laser lines projected and passed through the apexes of the first and fifth MPJ to confirm that axis of MPJ aligned to that of cradle on the transverse plane (Figure 3-8).

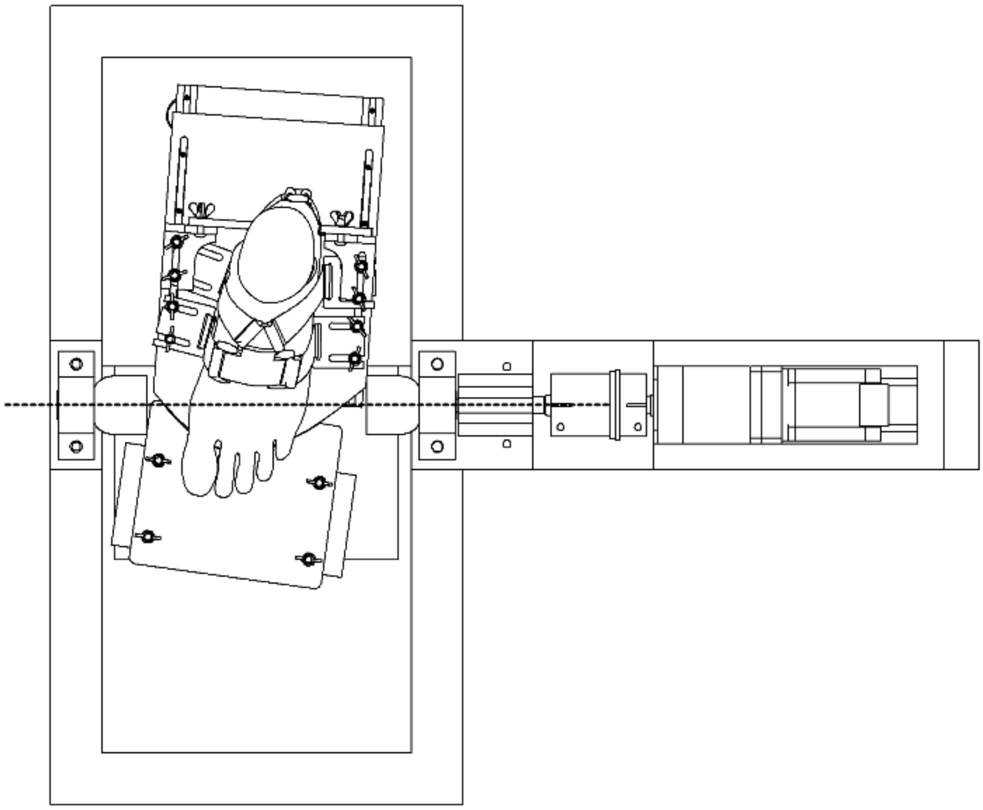


Figure 3-8 The foot platform was placed on the additional plate and toe platform was placed on the electromagnets of the cradle. After the first and fifth apexes of MPJ passed through the laser line projection, electromagnets were turned on to hold the platforms.

Afterwards, the electromagnets were turned on and held the platforms and foot in position. The cradle, toe platforms and toes were extended by the motor-sensor unit in predefined angular speed and displacement cyclically (Figure 3-9).

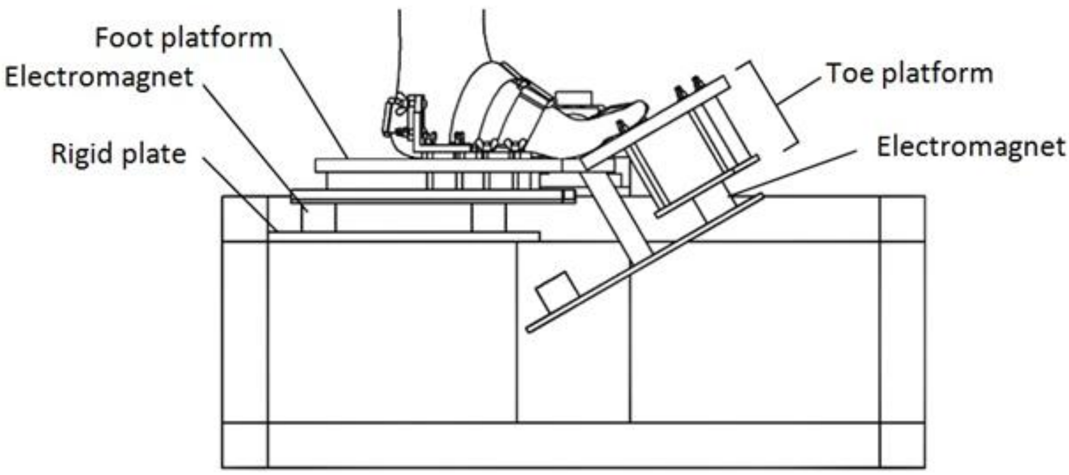
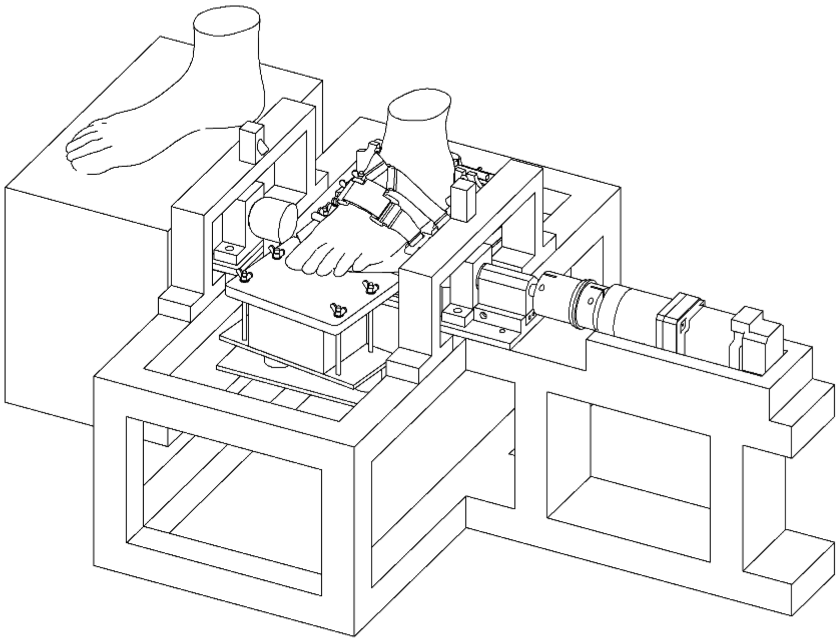


Figure 3-9 This figure illustrated how the toes were extended

At the same time, the torque transducer recorded the torque, T_1, which included the torque contributed by the cradle, the toe platform and MPJ. Next, the foot was removed from the foot platform and determine the torque, T_2, contributed by the cradle and the toe platform only. Subsequently the net MPJ torque (T_{MPJ}) was the torque difference in former and later condition, i.e.

$$T_{MPJ} = T_1 - T_2$$

The other foot was supported by a stand that the height was on the same level of the foot platform. Since the front and back sides of console were symmetric, the system could be applied to another side of foot by turning the subject to opposite direction (Figure 3-10).



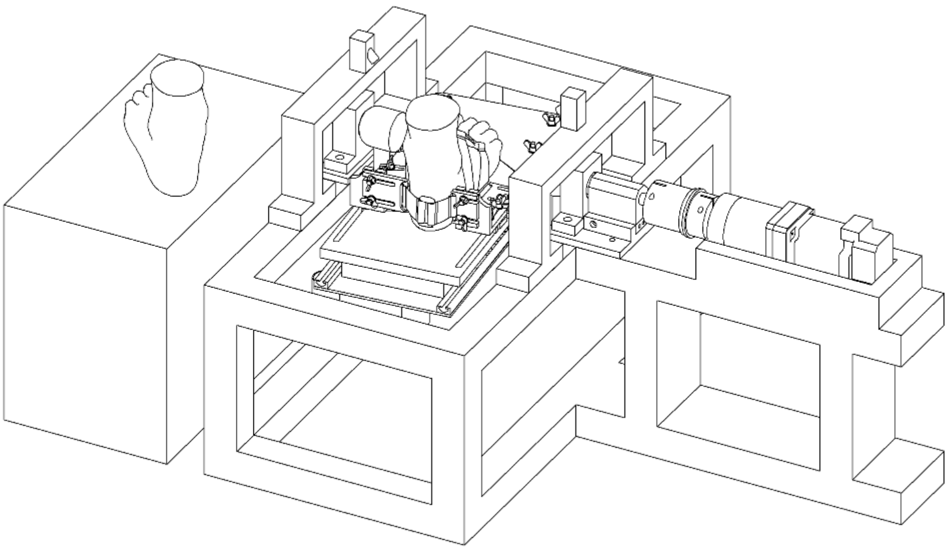


Figure 3-10 The dynamometer could measure MPJ of both sides of foot by turning the body to opposite direction.

Figure 3-11 illustrates how the dynamometer measure T_{MPJ} in standing position.

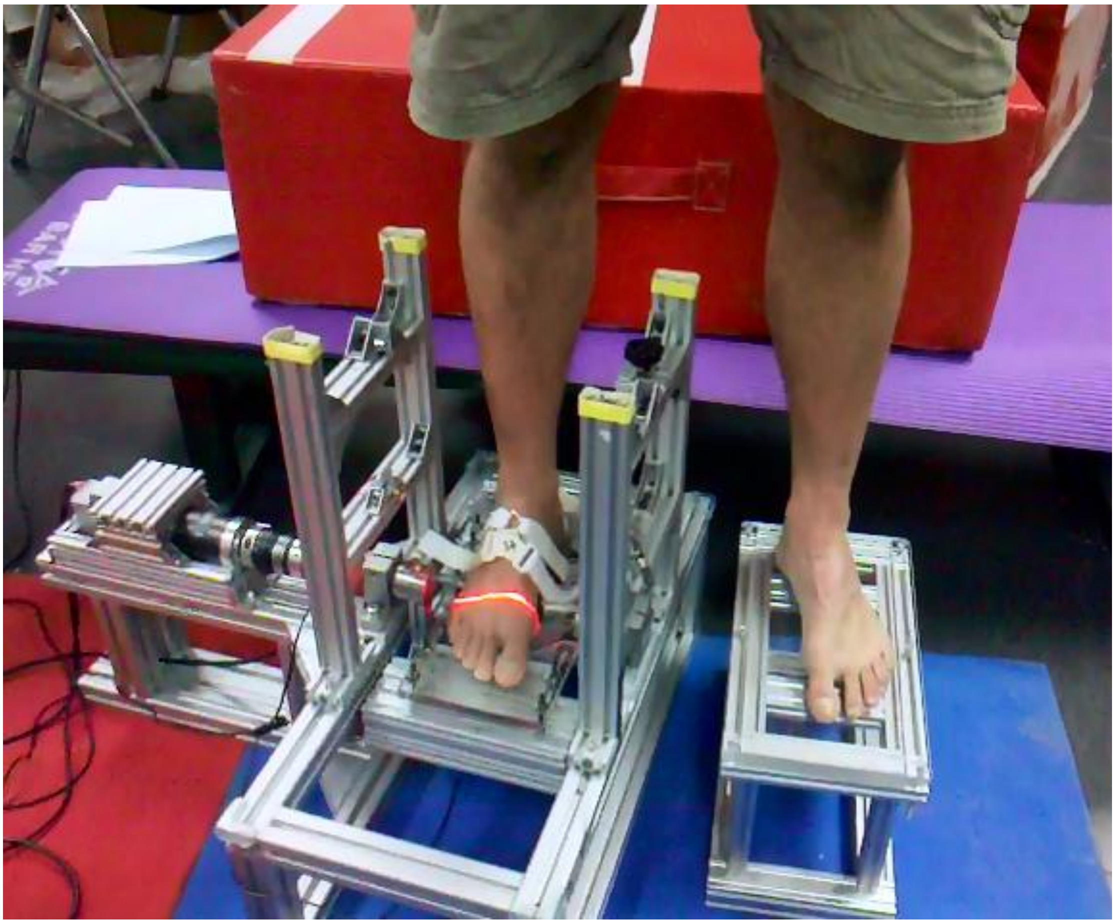


Figure 3-11 This photo illustrated how the the passive MPJ torque was measured in standing position

The procedures for ankle joint torque measurement were similar but simpler than that of MPJ configuration. The ankle joint axis was defined as a horizontal straight line passing through the apex of fibular malleolus and apex of tibia malleolus. Since inversion and eversion of ankle joint will respectively tighten lateral and medial ligaments of ankle joint and eventually increased the joint passive stiffness in plantar-flexion and dorsi-flexion, the ankle joint should be posited in natural position. The height of footplate was adjusted according to the aforementioned plate inserting method (Figure 3-4). The foot platform was moved on the cradle until the vertical laser lines projected on the apexes of tibia and fibular malleoli to make sure the axis of ankle joint aligned to that of cradle on the transverse plane (Figure 3-12).

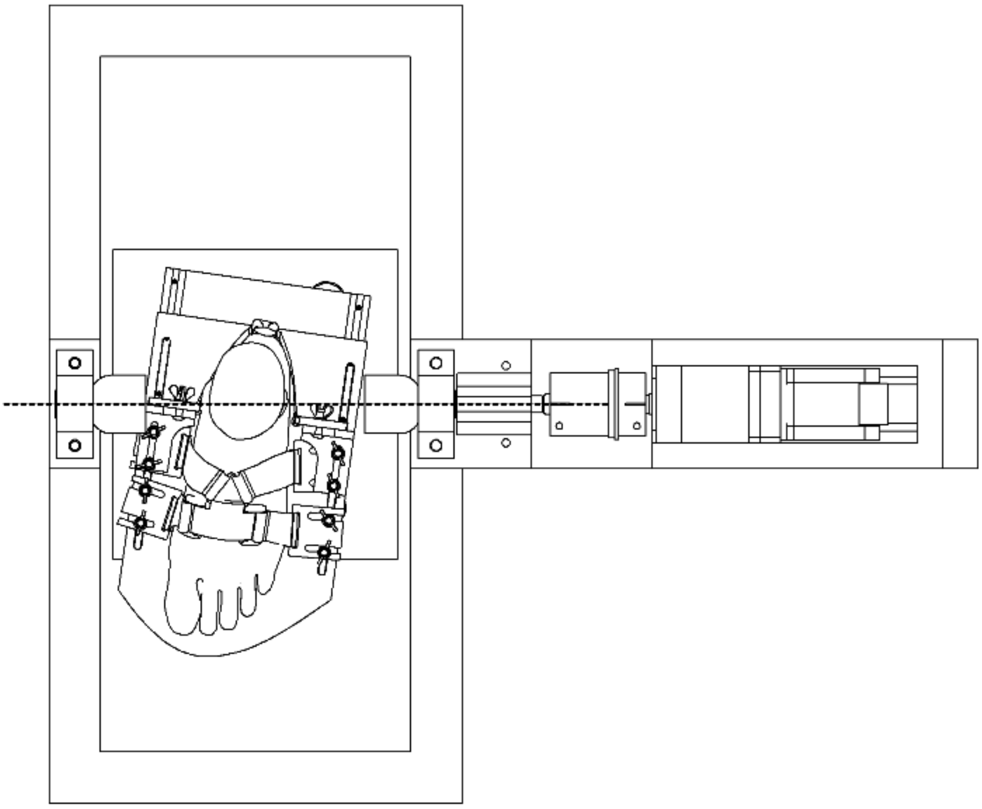


Figure 3-12 The foot platform could slide on the cradle to match the ankle joint alignment to the dynamometer on the transverse plane before the electromagnets were turned on

A rigid frame with matched height of foot platform was used to support another side of foot. Similar to MPJ measurement, the front and back sides of the console were symmetrical and the ankle joint torque of another side of leg could be measured by turning the body to opposite direction (Figure 3-13).

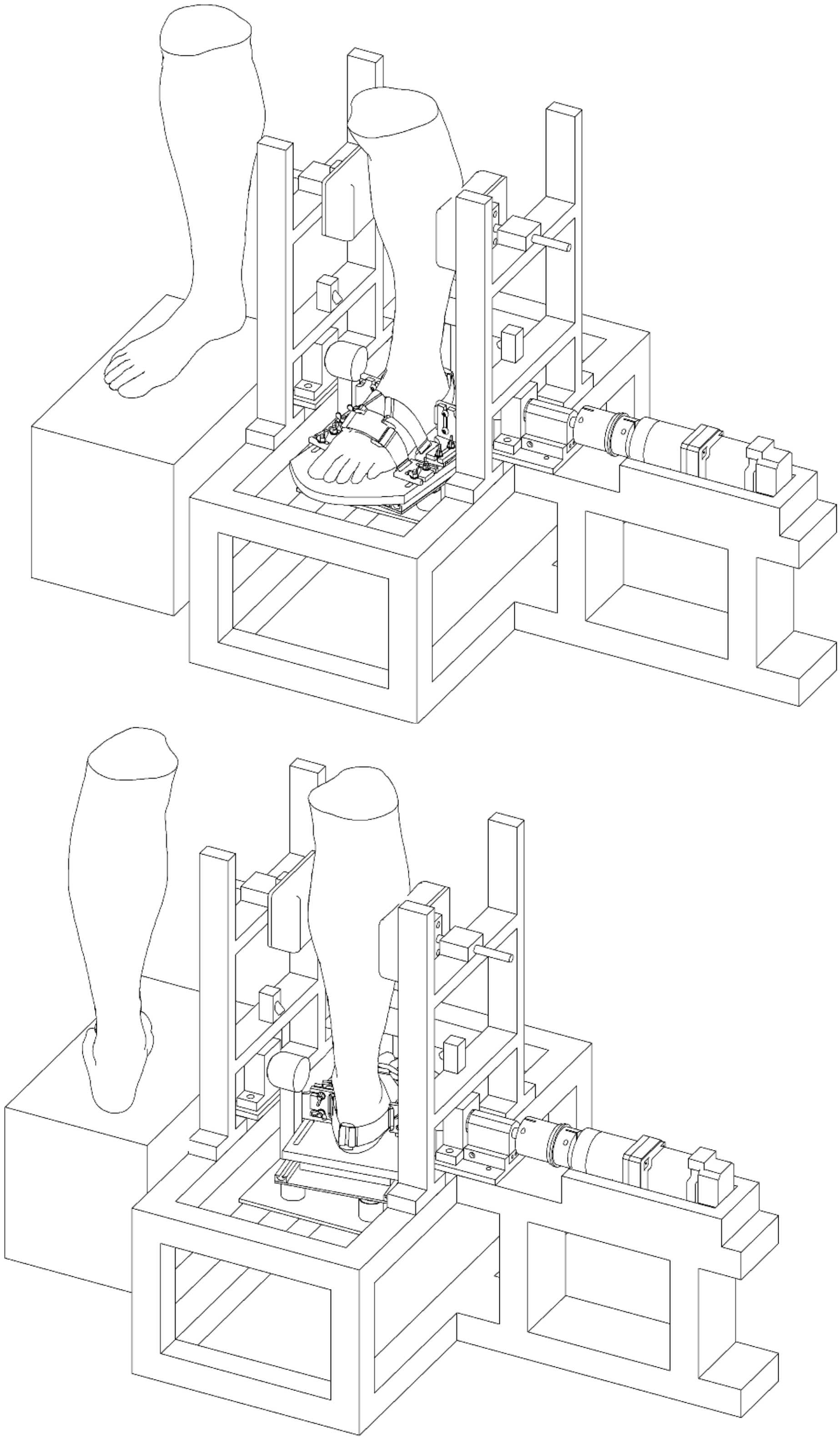


Figure 3-13 The dynamometer could measure ankle joint of both sides of foot by turning the body to opposite direction

After the electromagnets were turned on to fix the foot position, the foot platform and the foot were flexed and extended in predefined angular speed and displacement cyclically by the motor-sensor unit (Figure 3-14).

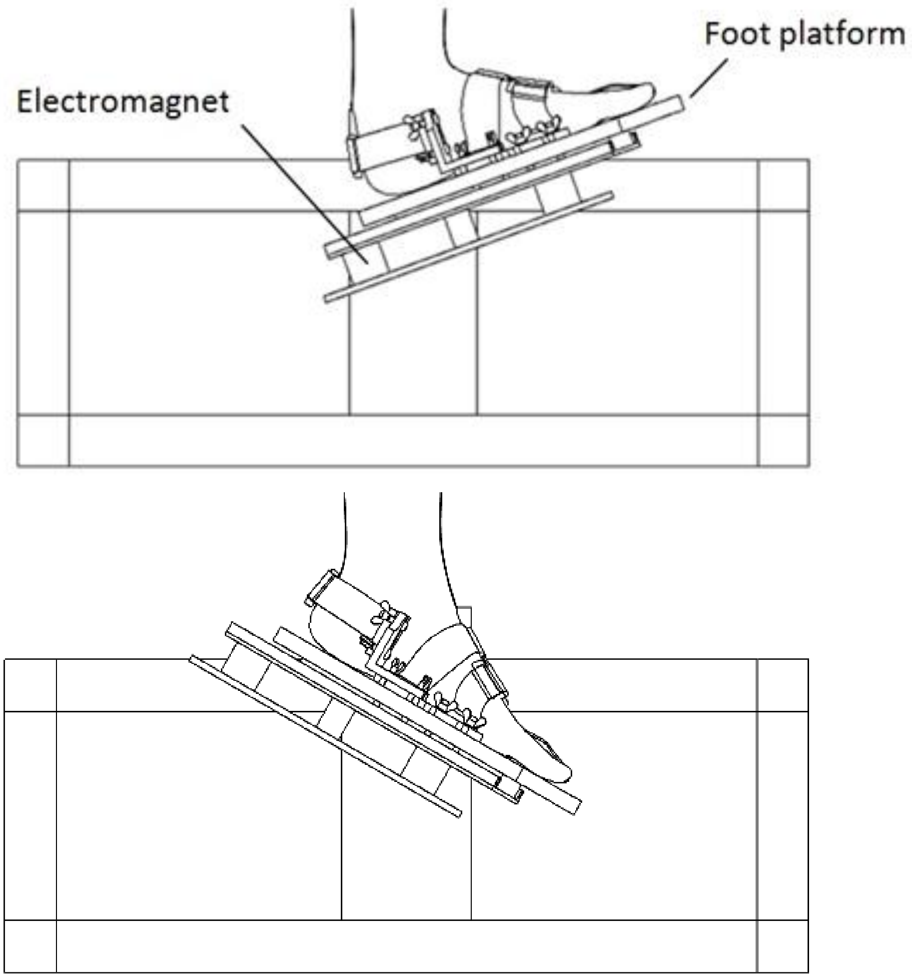


Figure 3-14 The ankle joint was flexed and extended by the dynamometer

The torque, T_A, contributed by the cradle, the foot platform and ankle joint was recorded. Similar to MPJ measurement, background torque, T_B, was determined in unloaded condition. The net ankle torque, T_{ankle}, was the subtraction of T_A and T_B, i.e.

$$T_{ankle} = T_A - T_B$$

Figure 3-15 illustrates how the dynamometer measures T_{ankle}.

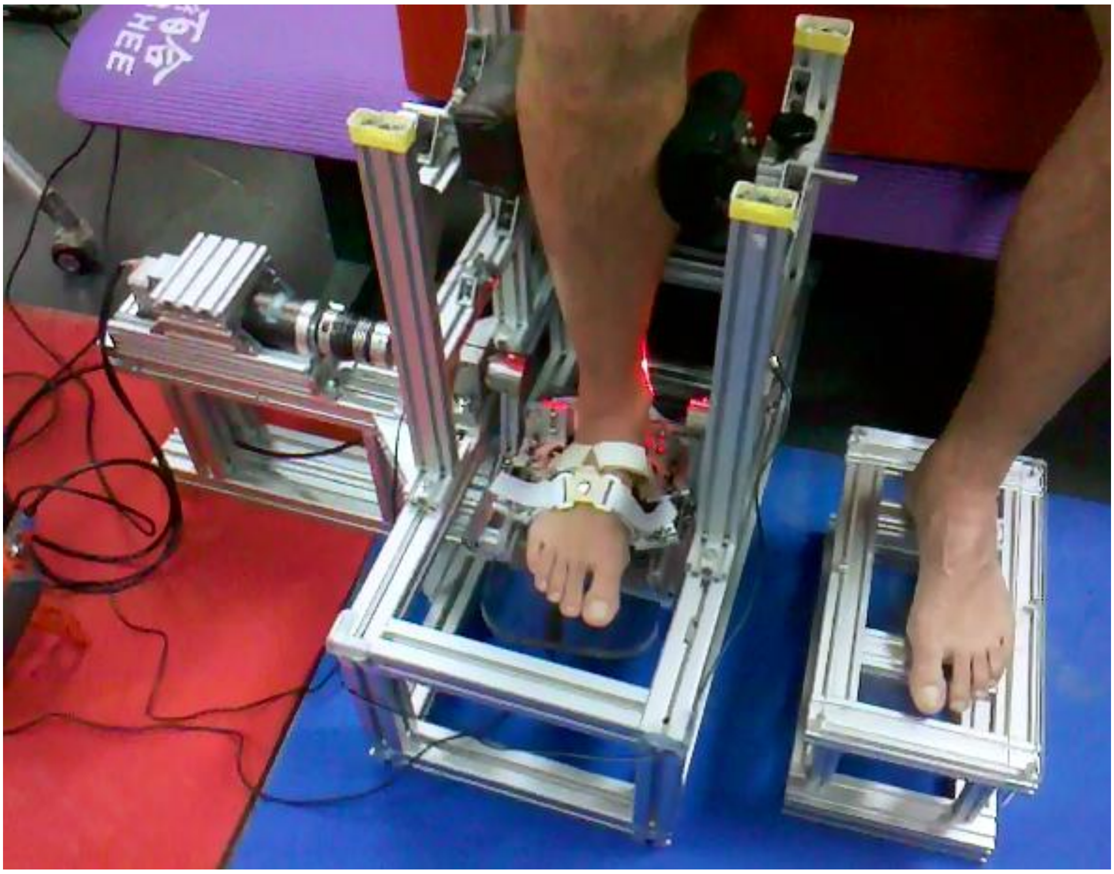


Figure 3-15 The photo illustrated how the passive ankle joint torque was measured by the dynamometer

Two set of devices were made as shown in Figure 3-16, one is arranged in MPJ configuration and another one was assembled in ankle joint configuration to eliminate the time consuming configuration-switchover steps and confirm both MPJ and ankle joint torque measurements were ready to start.

Figure 3-16 Two set of devices in different configurations prepared for measurement of both MPJ and ankle joint torque

3.3 Method

Within-day and between day reliability experiment were performed to test the reliability of the device.

3.3.1 Subjects

Nineteen male athletes (age: 23.1±3.4 years, height: 176.4±7.0 cm, weight : 74.2±9.1 kg, BMI: 23.8±2.4) participated in the reliability tests of the device. They were physically active undergraduate students with at least 6 hours exercise per week for at least 2 years so that the muscle strength reached a steady level All runners were free of any lower extremity injury in the six months prior to the start of the study to avoid the joint passive stiffness was contributed by formation of fibrous tissues rather than the strength of musculatures. Their foot sizes ranged from 6.5 to 13.5 (9.6±1.5) in US scale

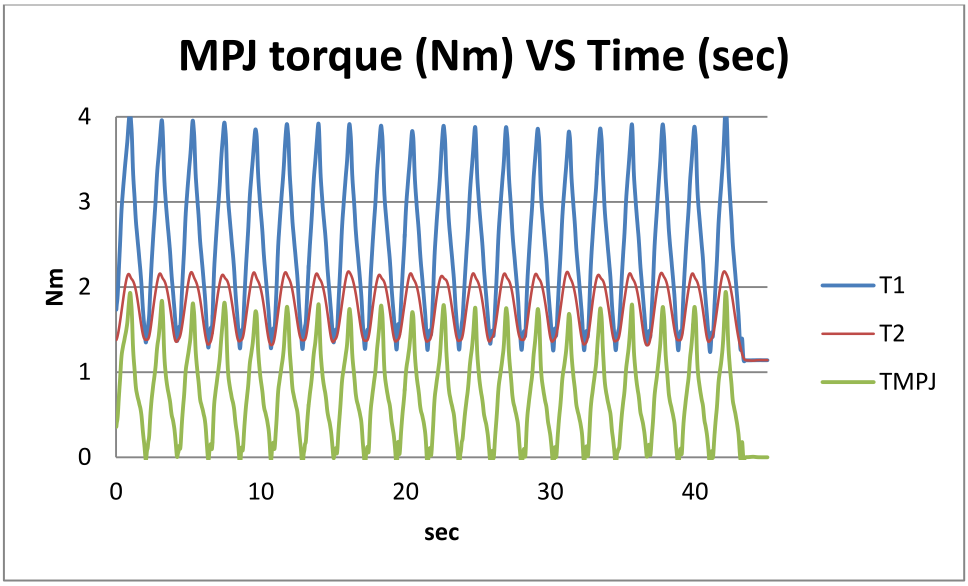
(The Brannock Device Co., Liverpool, NY, USA). All of them were physical education students of Beijing Sport University and regularly exercised for a minimum of six hours per week. This research was approved by the human subject ethics sub-committee of The Hong Kong Polytechnic University, and informed consent was signed by all participants prior to enrolment as shown in Appendix A and Appendix B. Nine participants took part in the within-day reliability experiment, and ten participants in the between day test-retest experiment. Five participants in each group were measured to determine the torque of ankle and MPJ of the right foot. The remaining participants in each group were measured for the left foot with identical procedures.

3.3.2 Ankle joint and MPJ torque measurements

For both T_{MPJ} and T_{ankle} measurements, participants seated on a height-adjustable bench so that ankle, knee and hip joint of both legs flexed in 90° as starting position. T_{MPJ} and T_{ankle} were determined separately, with each structure measured twice in each trial. For evaluation of within-day reliability, each joint was measured for two trials (two times for each) on the same day. For evaluation of between-day reliability, each joint torque was measured for two trials (two times for each) on two separate days within one week. All tests were performed by the same tester trained extensively for accurate device usage. Each joint was tested in random and non consecutive order. Prior to data collection each subject walked on a treadmill at a speed of 2m/s for 5 minutes, as a preconditioning exercise. During the measurements, the cradle of the dynamometer was swung for 20 cycles at an angular speed of 40°/s. The absolute values of joint movement were preferred as these ranges were appropriate to imitate the motion achieved during normal gait. For T_{MPJ} determination, the toes were cyclically dorsi-flexed by the cradle from 0°to 40° dorsiflexion as shown in Figure 3-9, comparable to 29.2° ± 6.9° dorsiflexion of the great toe in normal walking. (Turner et al., 2007). For T_{ankle} measurement, the whole foot was cyclically pushed from 20° plantarflexion to 20° dorsiflexion as shown in Figure 3-14, which was comparable to 18.5° ± 3.2° ankle joint range of motion for both plantarflexion and dorsiflexion during normal gait (Turner et al., 2007).

3.4 Data Analysis

As the aim of this experiment was to determine the reliability of present device, simpler peak torque rather than stiffness of MPJ and ankle joint was selected as a parameter to calculate ICC. For the same reason, the inertia of the foot and toes were neglected because the body segments of subjects did not change between the trials, and the results of the reliability would not be affected. Torque data of MPJ and ankle joint measurements, T_1, T_2, T_A and T_B, were imported to Matlab (Mathworks, version R2010b, Natick, MA, USA.) programme. Digital Butterworth low-pass filter with a cut-off frequency of 8Hz for T_1 and T_A, and 1.5Hz for T_2 and T_B was applied to remove electronic noise and mechanical noise. The high-frequency electronic noise is a natural character of all electronic circuits and its frequency should be over 8Hz. The applied low-pass filter could remove electronic noise. The main source of low-frequency mechanical noise of present device was the physical vibration when the swing cradle suddenly stopped rotation and immediately rotated in opposite direction. When the cradle was moving in loaded conditions, the weight of foot or toes effectively suppressed the vibration and the output signals were relatively smooth. However, the vibration was obvious when the cradle was swing in unloaded condition because of weight reduction. Hence, lower cut-off frequency of low pass filter was selected to remove the lower frequencies noise signals in the data collected in unloaded condition (T_2 and T_B). T_{MPJ} and T_{ankle} were determined by subtracting the loading torque (T_A and T_1) from the unloading torque (T_B and T_2) as denoted by the green, blue and red curves respectively in Figure 3-17. Peak torque values of the 6th to 15th cycles were averaged for each measurement. The first five cycles were not accounted for analysis to ensure the subjects had enough time to be accustomed the passive movement. Finally, T_{MPJ} and T_{ankle} values were equal to the mean values of the averaged torque in the first and second measurements. T_{MPJ} and T_{ankle} were imported to statistical software (IBM SPSS version 19, NY, USA) to calculate ICC with absolute agreement (two ways mixed model) with confidence interval of 95% to determine test-retest reliability of the device.



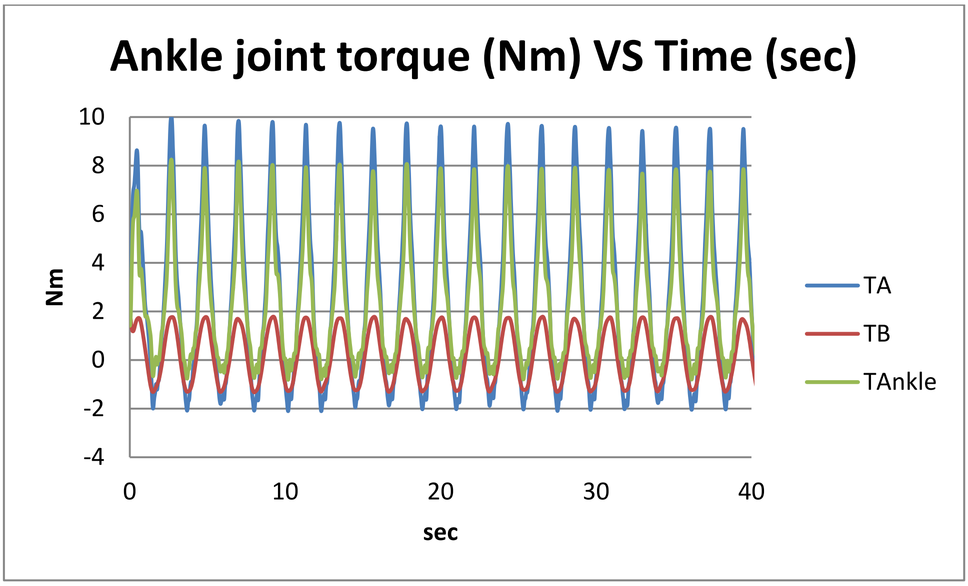


Figure 3-17 MPJ (above) and ankle joint (below) torque collected by the dynamometers. The net joint torque (green lines) were determined by the torque in loading condition (blue lines) minus the torque in unloading condition (red lines).

3.5 Results

Within-day and between-day reliability of T_{MPJ} and T_{ankle} measurements were presented as ICC scores. Individual T_{MPJ} and T_{ankle} are displayed in Table 3-1 and

Table 3-2.

Table 3-1 Individual T_{MPJ} and T_{ankle} of within-day repeatability test

	Joint torque (Nm)			
Participant	T_{MPJ} trial 1	T_{MPJ} trial 2	T_{ankle} trial 1	T_{ankle} trial 2
1	0.69	0.86	4.28	4.43
2	1.54	1.33	11.85	12.09
3	1.04	0.93	4.58	4.88
4	1.64	1.49	10.69	8.53
5	0.92	0.96	4.68	4.91
6	1.26	1.14	7.28	6.60
7	1.69	1.96	12.26	11.01
8	0.98	0.97	12.00	13.05
9	0.81	0.94	5.48	6.17
10	1.36	1.27	6.47	6.70

Table 3-2 Individual T_{MPJ} and T_{ankle} of between-day repeatability test

	Joint torque (Nm)			
Participant	T_{MPJ} day 1	T_{MPJ} day 2	T_{ankle} day 1	T_{ankle} day 2
1	1.62	1.51	10.97	11.72
2	1.95	2.27	10.55	10.01
3	1.72	1.44	7.74	8.29
4	1.43	1.30	7.65	6.34
5	1.08	1.06	4.83	4.28
6	1.53	1.66	6.91	7.49
7	1.21	1.18	6.03	6.05
8	1.09	1.00	9.89	10.16
9	2.00	2.02	10.02	9.40

Table 3-3 and Table 3-4 indicate that the device and procedures can provide highly repeatable measurements of T_{MPJ} and T_{ankle}.

Table 3-3 Mean T_{MPJ} and T_{ankle} and between-day reliability scores

	Mean (SD)	ICC	95% CI	p-value
T_{MPJ}	1.50 (0.38)	0.91	0.66-0.98	>0.01
T_{ankle}	8.24 (2.2)	0.96	0.83-0.99	>0.01

Table 3-4 Mean T_{MPJ} and T_{ankle} and within-day reliability scores

	Mean (SD)	ICC	95% CI	p-value
T_{MPJ}	1.19 (0.34)	0.91	0.67-0.98	>0.01
T_{ankle}	7.90(3.18)	0.96	0.85-0.99	>0.01

3.6 Discussion

Although the ankle joint and MPJ passive stiffness had already been investigated with different types of dynamometers, precise control of angular speed and alignment of the measured joints, which largely affects the reliability especially at the MPJ were not reported. To the best of our knowledge, the presented device was the first disclosed dynamometer which could measure both MPJ and ankle joint passive stiffness with precise control of angular speed and displacement. The ability of measuring MPJ passive stiffness is extensively useful for clinical evaluation for different kinds of surgical or conservative treatment deal with toe deformities including hallux valgux, claw toe and cross toe. Medication response of toe related disorders, such as toe arthritis, can also be monitored by the device. Apart from clinical application, passive MPJ stiffness is related to sport performance. It is believed that the both of the foot arch and leg muscle strain can store energy during deceleration and release the stored energy in the acceleration phase of the following step to reduce cost of walking or running (Alexander, 1987). The toe plantarflexor muscles help the energy conservation by partially support the foot arch and act as an elastic component of the foot (Alexander, 1987). Besides, toe plantarflexor muscles provide propulsion in the final stance phase to push the body forward (Mann and Hagy, 1979). Since the strength of muscle was positively correlated to the passive stiffness of corresponding joint, it was reasonable to hypothesize that the passive MPJ stiffness was positively correlated to running economy, and power in term of running and jumping.

In most cases of the 19 measured subjects, the T_{MPJ} was even lower than the effective measureable torque range (2.1 to 21 Nm) of the torque transducer. However, the measured torque included the effect of the pedal and foot or toe platform in both loaded and unloaded conditions. Since the torque contributed by the pedal and foot or toe platform properly exceeded 2 Nm, the net MPJ torque was thus still measured within the effective measureable torque range of the transducer.

The results showed that the averaged T_{MPJ} was only 1.34 ± 0.39 Nm. The low value could be easily overridden by errors due to misalignment of the MPJ and the rotating axis of pedal. Nevertheless, the ICC in the repeatability tests was very high (≥.91). The specific features of this newly developed dynamometer enhance the reliability by properly align the ankle and MPJ axis with the assistance of the laser line projection, and control the angular speed and displacement of the toe and ankle with the computerized motor.

Although the vertical laser lines only projected on the ankle or toe joints, the proximal part of the lower limb was also controlled since the subjects were required to sit properly so that the ankle, knee and hip were flexed in 90 degrees without thigh adduction/abduction. The experiment showed that the reliabilities of torque measurement were very high even the laser lines controlled only the alignment of the foot and that of proximal part of the leg was controlled by checking of the hip posture.

The results would be inaccurate by improper posture since body weight would be contributed to the rotating pedal and mistakenly increased the torque readings. However, according to torque VS time graph in every trial such as Figure 3-17, the net torque of the ankle or toe were about zero whenever the pedal rotated to starting position. This reflected that the ankle, knee and hip joints were all well controlled.

The computer-controlled device can be used to measure the torque at maximal voluntary isometric contractions. It can also provide feed-back to control the angular speed and displacement of the pedal for training of MPJ and ankle joint in rehabilitation.

The ankle stiffness of people with diabetes mellitus is significant different from that of healthy subjects. It has been suggested that the increased stiffness plays a significant role in ulceration and other pathologies of the foot and ankle. The dynamometer can be applied to monitor the degree of ankle stiffness and thus provide appropriate preventive measure for foot ulceration. Since the experiment was only applied on normal subjects, the method of position control may need modification for patients with reduced locomotion ability. Clinical application of this device should be examined for improvement.

3.7 Conclusion

This study introduced an innovative dynamometer and protocol for MPJ and ankle joint torque measurements with precise and reliable foot alignment, angular joint speed and displacement control. Within-day and between-day test-retest measurements on MPJ and ankle joint torque were conducted on ten and nine healthy male subjects respectively. The ICC of averaged peak torque of both joints in within-day and between-day test-retest experiments were in the range between 0.91 to 0.96, indicating that the innovative device and the protocol were systematic and reliable for multiple scientific and clinical applications.

CHAPTER 4. Investigation of ankle-foot passive joint stiffness and leg vertical stiffness during hopping exercise.

4.1 Introduction

The foot is the interface of body and the ground for standing, walking and running, and its structure essentially affects the performance of locomotion (Viale et al., 1998, Ogon et al., 1999, Nachbauer and Nigg, 1992). As aforementioned in previous chapters, a flatfoot incidence survey conducted by Rao and Joseph (1992) indicates that children who wear shoes were significantly ($p<0.001$) more likely to suffer from flatfoot compared with those who are barefoot or wearing minimum shoes such as sandal, and the authors believe the unshod condition enhances the development of foot muscles for arch support and hence the incidence of flatfoot in barefoot or minimal shoes condition are significantly lower compared to shod condition. The authors also suggested the foot arch is partially supported by toe flexor muscles. In respect with performance of locomotion, it is estimated that 17% of the mechanical energy generated per step is stored and recovered by the elastic structures of longitudinal and transverse arches of foot (Alexander et al., 1987). It is reported that runners with low foot arch exhibit decreased vertical stiffness compared with high foot arch runners (Williams et al., 2004). Stefanyshyn and Nigg (1998), Chelly and Denis (2001) and Butler et al.(2003) suggested that running speed might be increased by greater K_{vert} or knee joint stiffness. In this respect, the strength of toes not only provides power for propulsion, but partially maintains the arch of foot for energy conversation and increases K_{vert} to increase running speed as well. Moreover, the passive stiffness can be increased by eccentric exercise in both rats and human. Pousson et al. (1990) reported passive stiffness gets increased by training elbow flexors in human. Reich et al. (2000) found that the development of optimum length force and passive stiffness could be enhanced by eccentric-based exercise program in rats. Researchers (Reich et al., 2000, Lindstedt et al., 2002, LaStayo et al., 2003) suggested that eccentric exercise causes an increase in passive stiffness (due to the titin filament), which could improve athletic performance in running and jumping. Askling et al., (2003) also showed that eccentric exercise enhances sprinting and jumping performance. In addition, it was shown that by strengthening toe flexor muscles jumping height and distance can be increased significantly (Goldmann et al., 2013). Apart from MPJ passive stiffness, Farley and Morgenroth et al., (1999) measured vertical stiffness and active joint stiffnesses of hip, knee and ankle by inverse dynamics during hopping and found that vertical stiffness primarily depends on active joint stiffness of the ankle joint. As a result, it is reasonable to hypothesize that both ankle and MPJ passive stiffness are related to vertical stiffness or leg stiffness. On the contrary, there was a study comparing K_{vert} and K_{leg} among the three fastest runners on earth, and it was found that both of the stiffness of the champion, Usain Bolt, was significantly lower than the other two competitors (Taylor and Beneke, 2012). Brughelli and Cronin claimed that relatively low stiffness enhances the elastic energy storage and utilization during stretch shortening cycle.

In summary, the toe muscle strength partially supported foot arch structure (Rao and Joseph, 1992), and hence it is related to K_{vert} and performance in running exercise (Williams et al., 2004, Taylor and Beneke, 2012). Meanwhile, it was claimed by some researchers that passive stiffness of joint and physical performance could be improved by eccentrically training (Reich et al., 2000, Lindstedt et al., 2002, LaStayo et al., 2003). However, there were some contradicting opinions that low K_{vert} and K_{leg} could improve running performance by enhancing elastic energy conservation (Taylor and Beneke, 2012). So, it is a problem that is worth to discuss. The aim of this study is to investigate the relationship between passive joint stiffness of MPJ and ankle joint and K_{vert} in hopping. It was hypothesized that the ankle-foot passive stiffness would be positively correlated to the K_{vert} during hopping.

4.2 Method

4.2.1 Subjects

Ninety-nine students were all from Beijing Sport University, aged 22.5±2.4 with shoe size 9±1.8 in US scale. All participants signed an informed consent form as shown in appendix A and ethical approval was granted by human subject ethics sub-committee of The Hong Kong Polytechnic University prior to the start of the study. All the subjects finished their experiment in one single day visit.

Inclusive criteria included

(1) Male gender

(2) Physically active undergraduate students, with at least 6 hours exercise per week for at least 2 years so that the muscle strength reached a steady level

(3) Not suffered from lower limb injury within 6 months to avoid the joint passive stiffness was contributed by formation of fibrous tissues rather than

the strength of musculatures.

4.2.2 Passive ankle joint and MPJ torques measurement

The T_{ankle}, MPJ_{sit} and passive MPJ joint torque in standing position (MPJ_{stand}) of both legs were measured by two custom-made dynamometers and, as described in chapter 3, one was put in ankle joint configuration and another was put in MPJ configuration for convenience (Figure 3-16). The present dynamometer is different from those in previous studies (Goldmann et al., 2013, Finley et al., 2012). The alignment in transverse plane and height were adjustable so that the MPJ axis and ankle joint axis (the horizontal line passing through the apex of first MPJ and fibular mallelous respectively) could align to the rotating axis of the dynamometer by the guide of laser line projections. In addition, all the toes were dorisflexed at the same time, so it was different from the device developed by Rao et al., (2011) which only examined the big toe. The custom made device was used to measure passive torques of ankle joint and MPJ in two different configurations. Prior to data collection each subject walked on a treadmill at a speed of 2m/s for 5 minutes, as a preconditioning exercise. In ankle joint measurement configuration, the whole foot was fixed on a footplate with Velcro strip and then was attached to the console in alignment. In MPJ measurement configuration, only the foot palm was fixed on the footplate with Velcro strips and all the toes were stretched out of the anterior edge, in other words, it is only the toe plate that supported the toes. The cradle of the dynamometer was swung for 20 cycles at an angular speed of 40°/s. For MPJ_{sit} determination, the toes were cyclically flexed by the cradle from 0° (horizontal level) to 40° dorsiflexion in sitting position, the range of motion was greater than the 29.2° ± 6.9° dorsiflexion of the great toe in walking (Turner et al., 2007). During the process, the torque resistance, T_1, was recorded.(Abdelkrim et al., 2007) The foot was then released from the footplate which was still on the console by electromagnets. Next the console started to swing the empty footplate again to record the background torque, T_2. The net joint torque, MPJ_{sit}, was the subtraction of T_1 and T_2. The procedures of MPJ_{stand} measurement were identical to that of MPJ_{sit} excepting posture change from sitting to standing. For T_{ankle} measurement, the whole foot was cyclically pushed from 20° plantarflexion to 20° dorsiflexion in sitting position, which was comparable to 18.5° ± 3.2° ankle joint range of motion for both plantarflexion and dorsiflexion during walking (Turner et al., 2007). The angular speed of MPJ and ankle joint movement was 40°/s and equivalent to two seconds for a complete cycle of movement. Angular speed, filtering technique and calculation were all identical to the MPJ_{sit} measurement method. All the measurements were taken in two trials.

4.2.3 Data collection for vertical stiffness

The amplitude of vertical reaction force and oscillation of CoM during hopping were determined to calculate K_{vert}. Subject was first standing on the force platform (Kistler 9281, Kistler Instruments, Winterthur, Switzerland) in sampling frequency of 1000 Hz in the order of right leg, left leg and both legs. They hopped with their hands putting on their hips and the frequency remained at 2.2Hz noticed by the sound of metronome for 20 seconds. Having a rest for 3 minutes, the subjects hopped again to finish the reminding conditions.

4.3 Data Analysis

4.3.1 Data analysis for passive ankle joint and MPJ torques

The data analysis techniques for T_{ankle}, MPJ_{sit} and MPJ_{stand} were identical. Fixed on the foot platform and swung to collect torque resistance T_A, then the foot was released from the dynamometer to measure the background torque T_B. Both the torque signals were transmitted to Matlab (Mathworks, version R2010b, Natick, MA, USA.) program and smoothened by 4th order zero-lag low-pass Butterworth digital filter with cut-off frequency 8Hz for T_A and 1.5Hz for T_B to remove noise signal as explained in section 3.2.3. The net joint torque signal during the motion was equal to T_A - T_B, and the joint torque was defined as the average values of peak joint torques in the middle ten of total twenty cycles (i.e. 6^{th} to 15^{th}).

4.3.2 Data analysis for vertical stiffness

For K_{vert} analysis, it could be referred to Cavagna G (1975) method (See section 2.3.1). The raw force-time data were imported to Matlab (Mathworks, version R2010b, Natick, MA, USA.) program and smoothened by 4th order zero-lag low-pass Butterworth digital filter with cut-off frequency 10Hz to remove noise signal. Since the hopping frequency was 2.2Hz, cut-off frequency of 10Hz was high enough to retain most signal components originated from the hopping movement but remove all unwanted high frequency signal. Body weight (BW) was first determined by force platform when the subject standing on it still, and then the body mass (m) of subject was equal to BW/9.8 automatically determined by the Matlab program. It eliminated the weight measurement procedure with other equipment and data input process to the software. The a_z was determined by F_z minus BW and divided by m ,i.e. a_z=(F-BW)/m. The 11th to 30th acceleration curves in each trials were independent from each other for calculation. Since the subjects hopped at 2.2 Hz for 20 seconds in each condition, there were about 40 acceleration curves collected in each trial. The 11th to 30th curves represented the middle section of data. Each acceleration curve was integrated over time to get the velocity (v'), i.e.

$$v' = \int a\,dt + C$$

Since the jumping height in the hopping testing was about the same in average, the constant C was assumed to be zero. And hopping velocity (v') in upward and downward movements were equal over time, the velocity for continuous hopping was equal to v' minus averaged v', i.e.

$$v = v' - \overline{v'}$$

The displacement of vertical CoM (Δz) was determined by the integration of v over time i.e.

$$\Delta z = \int v\,dt + C$$

Again, since the jumping height in the hopping testing was about the same in average, the constant C was assumed to be zero. Eventually, K_{vert} = maximum F_z/ peak Δy of averaged 11^{th} to 30^{th} hop cycles.

4.4 Results

4.4.1 Results for passive ankle joint and MPJ torques

The Mean and standard deviation of passive ankle joint and MPJ torques for both sides of feet were summarized in Table 4-1. The T_{ankle} was predictably higher than MPJ torques, because of large size of calf muscle and strong Achilles tendon. And it was stiffer than MPJ_{sit} and MPJ_{stand} by about 6 and 2.5 folds respectively. The MPJ_{stand} was about 2.5 times higher than MPJ_{sit}.

Table 4-1 Averaged and standard deviation of passive joint torque of ankle and MPJ in (Unit in Nm)

	Left			Right		
	T_{ankle}	MPJ_{sit}	MPJ_{stand}	T_{ankle}	MPJ_{sit}	MPJ_{stand}
Mean	8.43	1.43	3.46	8.29	1.42	3.54
SD	2.55	0.45	1.13	2.50	0.45	1.23

Pearson's correlation analysis was employed to test the correlations of T_{ankle}, MPJ_{sit} and MPJ_{stand}, and all the results were significantly correlated ($p<0.001$) and tabulated inTable 4-2. The correlation coefficients were ranged from 0.41 to 0.72.

Table 4-2 Pearson's correlation of T_{ankle}, MPJ_{sit}, and MPJ_{stand}

	Left		Right	
	p-value	R	p-value	R
T_{ankle} - MPJ_{sit}	**<0.001**	**0.64**	**<0.001**	**0.61**
T_{ankle} - MPJ_{stand}	**<0.001**	**0.41**	**<0.001**	**0.51**
MPJ_{sit} - MPJ_{stand}	**<0.001**	**0.70**	**<0.001**	**0.72**

4.4.2 Results for vertical stiffness

The mean and standard deviation of K_{vert} were tabulated in Table 4-3. The K_{vert} of left and right sides of legs were similar and closed to 10.1kN/m. For both legged hopping, the K_{vert} was 14.61 kN/m and approximating higher than that of single leg hopping by half. Table 4-4 showed the Pearson's correlation of the K_{vert} in different hopping conditions.

Table 4-3 Averaged and standard deviation of K_{vert} of different hopping conditions

	Vertical stiffness (kN/m)		
	Both	Left	Right
Mean	14.61	10.15	10.08
SD	3.20	1.72	1.77

Table 4-4 Pearson's correlation of the K_{vert} of different sides of legs

	p-value	R
Both - Left	**<0.001**	**0.64**
Both - Right	**<0.001**	**0.69**
Left - Right	**<0.001**	**0.88**

4.4.3 Correlation of ankle-foot passive joint torques and vertical stiffness

Pearson's correlation was used to identify the correlations of ankle-foot passive joint stiffness and K_{vert}. T_{ankle}, MPJ_{sit}, and MPJ_{stand} were all significantly correlated to K_{vert} for both sides of legs. However, the correlation coefficients were only low to moderate, ranged from 0.29 to 0.43.

Table 4-5 Pearson's correlation of the K_{vert} and passive joint torques

	Left K_{vert}		Right K_{vert}	
	p-value	R	p-value	R
T_{ankle}	**>0.001**	**0.43**	**>0.001**	**0.41**
MPJ_{sit}	**>0.001**	**0.37**	**>0.001**	**0.39**
MPJ_{stand}	**0.004**	**0.29**	**0.001**	**0.33**

4.5 Discussion

None of the previous researches had determined the T_{ankle} by dynamometer in range of 20° plantar and dorsi flexion in angular speed 40°/s in sitting position. It was difficult to determine the accuracy of the present dynamometers, although the ICC of within day and between day repeatability over 0.9 as shown in Chapter 3. In Table 4-2, it shows that the T_{ankle}, MPJ_{sit} and MPJ_{stand} are significantly, but only moderately, correlated. It showed that the stiffness of the joint is only partially dependent, and there are some other independent factors that affect stiffness. For example, the plantar-flexion of ankle and MPJ highly depend on independent Achilles tendon and plantar aponeurosis respectively. Even for same MPJ joint, the torque changed significantly in different postures (R = 0.7). It could because the foot arch was pressed down and the plantar aponeurosis was tightened by body weight in standing position according to windlass effect (Hicks, 1954). These factors reduced the value of correlation coefficient.

In Table 4-3, the K_{leg} of individual side is lower than that of both legs by about 40%. It may attribute to the reason that a single leg has to sustain double pressure compared with both leg hopping. The vertical stiffness acquired when both leg hopping was very comparable to the finding of two researchers, Farley and Morgenroth et al., (1999), who claimed that the vertical stiffness was 14.5kN/m (SD=0.7) in same hopping frequency and preferred height. In Table 4-4, the K_{vert} of left and right legs were highly correlated. It indicated that the leg stiffness does not depend on dominant side of human being.

In Table 4-5, K_{vert} is significantly and positively correlated with passive joint stiffness for both ankle and MPJ of corresponding sides, because the strong muscle groups of ankle and MPJ not only increase the passive joint stiffness, but also strengthen the leg to resist collapse during hopping. In addition, it was reported that K_{vert} mainly depends on active ankle joint stiffness during hopping (Farley and Morgenroth, 1999). However, the correlation of ankle joint stronger than that of MPJ has not been explained.

The measured joint stiffness may be due to formation of fibrous tissues because of injury rather than the strength of musculatures. So, the exclusive criteria of subject recruitment included free of lower limb injury prior to six months of the experiment and large amount of subject was hired to reduce the effect of joint passive stiffness contributed by formation of fibrous tissues.

4.6 Conclusion

The results support the hypothesis that ankle-foot passive stiffness is correlated to K_{vert} in hopping ($p>0.005$), however, the correlation is relatively weak ($R=0.29-0.43$). When the subjects were hopping, only the balls of the toes touched the ground and their heels were completely left up. It is assumed that the strength of the toe flexors is the main determinant of vertical stiffness, and hence the passive MPJ torque would be moderately or strongly correlated with vertical stiffness.

CHAPTER 5. Comparison of Ankle foot passive stiffness and vertical leg stiffness of marathon runners, basketball players and miscellaneous group of athletes

5.1 Study Summary

Achilles tendon of ankle and plantar aponeurosis of toes get short in the process of plantarflexion to store elastic energy, which will be released in the process of stretching in dorsiflexion when people running, jumping and walking. Besides individual joints, the whole leg is regarded as a spring to store elastic energy. It is thought that passive MPJ and ankle joint torques and K_{vert} are related to running economy, especially important to marathon runners. Besides, toes provide important function to body balance, jumping and propulsion, which are vital for basketball players. We hypothesized that marathon runners need to develop stiffer T_{ankle} to store more elastic energy for better running economy, and basketball players need to have stiffer toes for stronger toe plantar flexor development, which is important for their performance. Fifteen marathon runners, 19 basketball players and 17 miscellaneous athletes were recruited in the experiment. The custom-made dynamometers were used for both ankle and T_{MPJ}(sit) and T_{MPJ}(stand) measurements with controlled alignment and rotating speed. K_{vert} was determined by force platform and double integration method. One-way ANOVA was used to test the difference between the two groups, and Post-hoc test showed T_{ankle} of basketball players is significantly stiffer than that of miscellaneous athletes by about 24% ($p=0.03$). T_{MPJ}(sit) and T_{MPJ}(stand) of basketball players are stiffer than that of marathon runners and miscellaneous athletes by about 24% and 32% respectively ($p=0.01$). The results supported the first hypothesis that the toe plantar flexors of basketball players were stronger but not supported the second hypothesis that the ankle joints of marathon runners were stiffer for better RE. This study concluded that strengthening both ankle and toe plantar-flexors could improve basketball players’ performance.

5.2 Introduction

Ankle joint is the largest joint in foot, which connects foot and leg. Basically, it is hinge joint that allows foot dorsiflexion and plantarflexion. Calcaneus of ankle joint connects medial and lateral gastrocnemius muscles via large Achilles tendon which provides foot powerful plantarflexion for propulsion. Achilles tendon in ankle gets shorter during plantarflexion to store elastic energy and then this energy will be released when people stretching in dorsiflexion (Ishikawa et al., 2005a) in running (Roberts et al., 1997) and jumping (Ishikawa et al., 2005b). The T_{ankle} is thought to be related to running economy (Lichtwark and Wilson, 2008).

MPJ is the second largest joint in foot. It connects toes that are the most distal part of lower extremities and play an important role in locomotion. During walking, the toes exert pressures on the ground for about three quarters of the standing phase and share the pressure with metatarsal heads. It plays significant function for enlarging the weight-bearing region in walking (Hughes et al., 1990). Apart from reducing plantar pressure, the toes maintain the body balance. Mueller et al. (1997) found that trans-metatarsal amputees could not maintain balance as well as aged-matched controls. Touching ground, the toes keep body balance in different postures (Tanaka et al., 1996). Conversely, it is evident that weak toe flexors will not only cause both toe deformities and functional abnormalities (de Win et al., 2002, Senda et al., 1999), but also increase the risk of tumbling for senior citizens (Mickle et al., 2009). Apart from balance, sufficient toe flexor strength is necessary to change direction, provide propelling force at starting stage and reduce shock during landing (Mann and Hagy, 1979, Senda et al., 1999). Goldmann et al.,(2013) custom made a dynamometer for toe maximal voluntary isometric contraction training. After 7 weeks, the jumping height and distance significantly increased. The authors suggested "The increased force potential made a contribution to an athlete's performance enhancement". However, athletes' performance enhancement also includes endurance. When the toes were extended, the plantar aponeurosis, which inserted from the base of calcaneum to the phalanxes via base of MPJ, become tightened (Hicks, 1954). During the shortening phase of the stretch-shortening cycle, the plantar aponeurosis together with other toe flexors muscles may reduce energy cost in submaximal running by returning elastic energy (Jones, 2002). In fact, some other evidence indicates musculotendinous stiffness negatively correlates with the energetic cost of treadmill running (Dalleau et al., 1998). Metabolic and mechanical energy generated from muscles will be stored as elastic energy and then be returned to reduce the energy cost of human locomotion (Hof et al., 1983, Alexander, 1991).

In addition to particular joints, it is also believed that energy will be stored and released in the shortening and stretching cycle of whole leg during locomotion.

K_{vert} describes vertical motion of the CoM during contact phase, and the resistance of an object or a body in length change is described as stiffness (McMahon and Cheng, 1990, Cavagna et al., 1991, Seyfarth et al., 2002), which is often applied to animal and human lower extremities to describe mechanical behaviors. During typical human movement such as running and jumping, elastic energy alternatively stores and releases in musculo-tendinous unit such as muscles, ligaments and tendons of leg. The leg and foot store elastic energy, and then recoil in the second half of stance phase in order to push the body upwards and forwards. Since the mechanical stiffness of human legs can be determined during normal human movement and it has a key influence on functional performance, it has been an area of interest for researchers for many years. It was reported that higher leg vertical stiffness reduces energy cost in sub-maximal running (Dalleau et al., 1998).

In basketball games, players run around the court in different directions with sudden stopping, jumping, and maintaining body balance, and all the actions require strong toe flexors to be achieved. In addition, marathon runners can perform better by reducing the energy cost, which may relate to MPJ and ankle joint stiffness and vertical stiffness. In this study, it is hypothesized that 1) the Passive MPJ torque of basketball players is significantly higher than marathon runners and miscellaneous athletes due to high demand of toe function and 2) the joint passive stiffness and vertical stiffness during hopping of marathon runners are significantly higher than other two groups.

5.3 Methods

5.3.1 Subjects

Fifty-one male students aged 22.5±2.4 were from Beijing Sport University. There were 15 marathon runners, 19 basketball players, and 17 miscellaneous athletes who regularly participated in different kinds of sports such as table tennis, badminton and football etc. Their shoes sizes were ranged from 6 to 10 in US scale. All participants signed an informed consent form as shown in appendix A and ethical approval was granted by human subject ethics sub-committee of The Hong Kong Polytechnic University prior to the start of the study. All the subjects finished their experiment in one single day visit.

5.3.2 Ankle joint and MPJ passive joint torque measurement

T_{ankle}, MPJ_{sit} and MPJ_{stand} measurements were introduced in chapter 3. The dynamometers adopted in this experiment were different from those in previous studies (Rao et al., 2011, Senda et al., 1999). The alignment in transverse plane and height were adjustable so that the MPJ axis and ankle joint axis (the horizontal line passing through the apex of first MPJ and fibular mallelous respectively) can align to the rotating axis of the dynamometer by laser line projection guide. In addition, all the toes, rather than only the big toe, were dorisflexed by electric motor in controlled speed and angular displacement. The custom made device was used to measure T_{ankle}, MPJ_{sit} and MPJ_{stand} in two different configurations. In ankle joint measurement configuration, the whole foot was fixed on a footplate by Velcro strips and then was attached to the console in alignment. The foot flexed and extended in +20° to -20° in relaxation with angular speed 40°/s for 20 cycles in sitting position.

Meanwhile, the torque resistance, T_A, was recorded. The foot was then released from the footplate while it was still attached to the console by electromagnets. The console swung the empty footplate again to record the background torque, T_B. Both the torque signals were smoothened by 4th order zero-lag low-pass Butterworth digital filter with cut-off frequency 8Hz for T_A and 1.5Hz for T_B to remove noise signal as explained in section 3.2.3. The net ankle torque sequence, T_{ankle}, was equal to T_A - T_B, and the ankle torque was defined as the peak of T_{ankle} in the middle 10 cycles.

In MPJ measurement configuration, only the foot palm was fixed on a footplate by Velcro strips, and all the toes stretched out of the anterior edge. Only the plate supporting the toes was swinging by the consoles. The toes were dorsi-flexed from horizontal level to 40° dorisflexion for 20 cycles. Angular speed, filtering technique and calculation were all identical to the ankle joint measurement method. However, in the process of MPJ torque measurement, both sitting and standing position were taken in two trials. Both left and right sides of the joints were measured, and the overall joint torque was the averaged value of both sides to prevent data duplication (Menz, 2005).

5.3.3 Vertical stiffness measurement

For K_{vert} measurement, it referred to Cavagna G method (1975). Subjects were first standing on a force platform in the order of right leg, left leg and both legs. They hopped with their hands putting on the hips at a frequency of 2.2Hz noticed by the sound of metronome for 20 seconds. After taking a rest for 3 minutes, the subjects hopped again with the reminding conditions. The raw ground reaction force (F) was smoothened by low-pass 4th order digital zero-lag Butterworth filter with cut-off frequency 10Hz to remove noise signal. The a_z was then determined, and it was equivalent to F minus BW and then divided by body mass ,i.e. a_z=(F-BW)/m. 11th to 30th acceleration curves were isolated from each other for double integration to get Δ y, d. K_{vert} was defined as the maximum reaction force divided by the maximum displacement, i.e.

$$\mathrm{K_{vert}} = \frac{\mathrm{F_{z(max)}}}{\Delta \mathrm{y_{(max)}}}$$

5.4 Data Analysis

The statistics of the passive joint torques and K_{vert} were imported to statistical software (IBM SPSS version 19, NY, USA) to be analyzed with one-way ANOVA followed by Post-hoc test (Bonferroni) to test significant differences among the groups (p=0.05).

5.5 Results

Table 5-1 shows the mean and standard deviation of ankle joint and MPJ torques for the 3 groups of subjects.

Table 5-1The average and standard deviation of passive joint torques (Nm).

	T_{ankle}	MPJ_{sit}	MPJ_{stand}
Marathon	7.37 (1.80)	1.24 (0.39)	2.96 (0.84)
Basketball	8.67 (2.41)	1.63 (0.46)	3.85 (0.88)
Miscellaneous	6.98 (1.27)	1.22 (0.25)	2.93 (0.72)

Table 5-2 shows the mean and standard deviation of vertical stiffness in different hopping conditions for the 3 groups of subjects.

Table 5-2The average and standard deviation of vertical stiffness (kNm/m).

	K_{vert} Both	K_{vert} Left	K_{vert} Right
Marathon	14.40 (2.56)	9.79 (1.70)	9.46 (1.59)
Basketball	15.08 (3.01)	9.99 (1.73)	9.88 (1.67)
Miscellaneous	14.34 (2.30)	9.80 (1.25)	9.86 (1.53)

Table 5-3 shows no significant difference between groups were found in K_{vert} for all hopping conditions ($P>0.05$). However, significant differences are found in T_{ankle} ($p=0.03$), MPJ_{sit} ($p<0.01$) and MPJ_{stand} ($p<0.01$).

Table 5-3 ANOVA according to different parameters, significant differences were marked by '*'.

Parameter	Mean square	F	p
T_{ankle}*	14.09	3.85	0.03
MPJ_{sit}*	0.93	6.77	>0.01
MPJ_{stand}*	4.94	7.33	>0.01
K_{vert} Both	3.06	0.43	0.65
K_{vert} Left	0.23	0.09	0.91
K_{vert} Right	0.90	0.35	0.71

In Table 5-4, post hoc test (Bonferroni) indicates T_{ankle} of basketball players is not significantly stiffer than that of marathon runners (p=0.17), but significantly stiffer than that of miscellaneous athletes by about 24% (p=0.03). MPJ_{sit} and MPJ_{stand} of basketball players are stiffer than that of marathon runners and miscellaneous athletes by about 24% and 32% respectively (p=0.01). The results are summarized in Figure 5-1 to Figure 5-3.

Table 5-4 Results of Post hoc test (Bonferroni) of passive MPJand ankle joint torques according to different types of sport, significant difference were marked by '*'

			Mean difference (Nm)	Percentage	p-value
T_{ankle}	Marathon	Basketball	-1.30	-15%	0.17
	Basketball	Miscellaneous	1.69*	24%	0.03
MPJ_{sit}	Marathon	Basketball	-0.40*	-24%	0.01
	Basketball	Miscellaneous	0.41*	33%	0.01
MPJ_{stand}	Marathon	Basketball	-0.89*	-23%	>0.01
	Basketball	Miscellaneous	0.92*	31%	>0.01

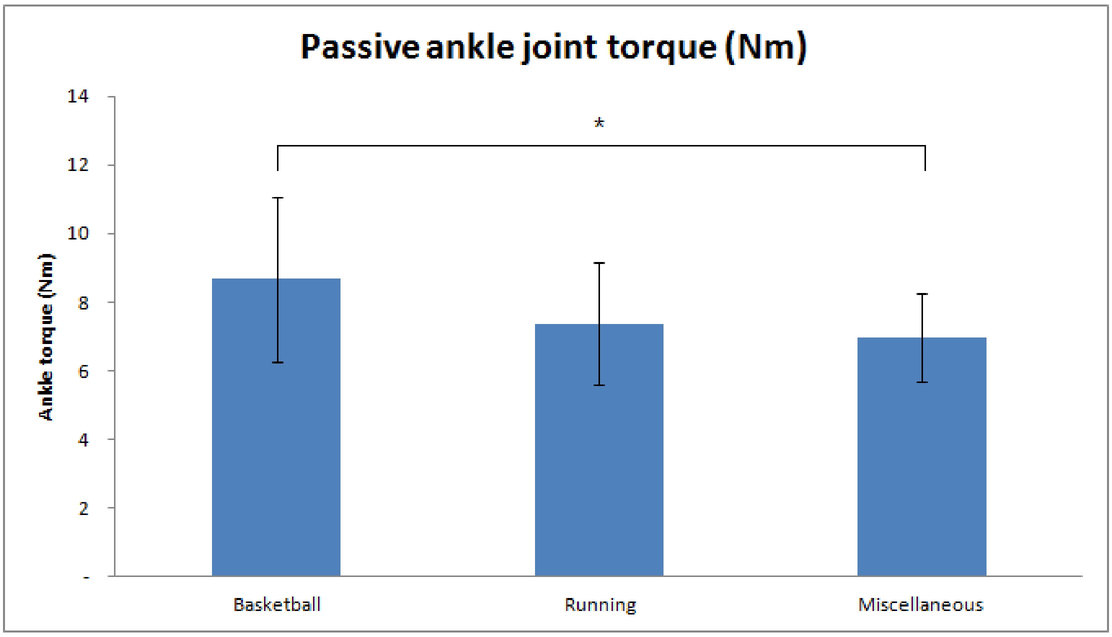


Figure 5-1 T_{ankle} (Nm) among marathon runners, basketball players and miscellaneous athletes. T_{ankle} of basketball players was significantly higher than miscellaneous group of athletes.

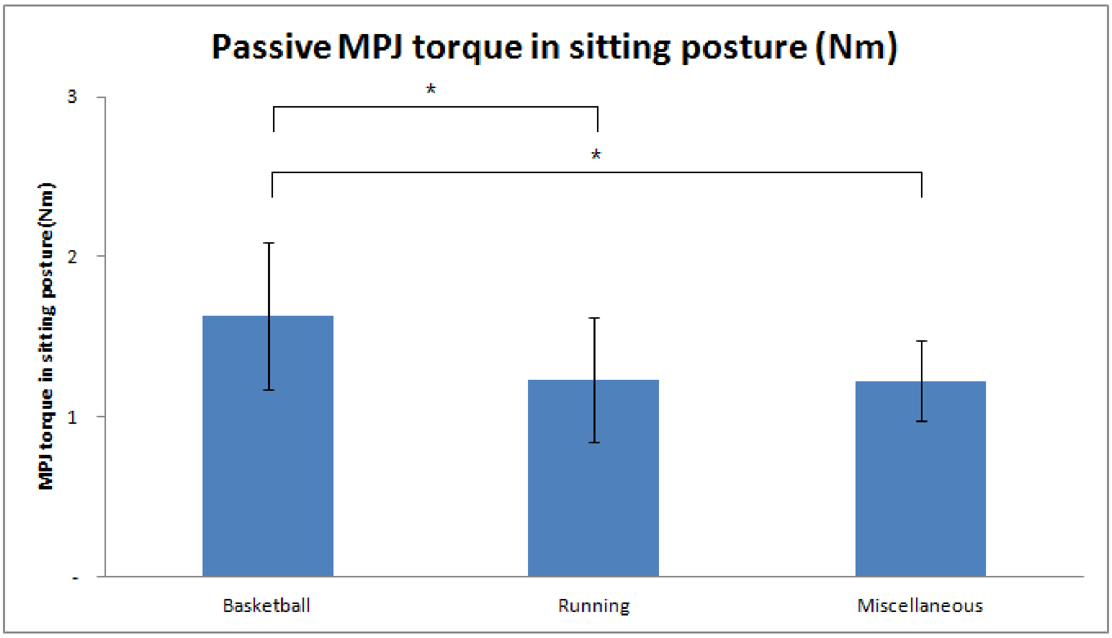


Figure 5-2 MPJ_{Sit} (Nm) among marathon runners, basketball players and miscellaneous athletes. MPJ_{sit} of basketball players was significantly higher than the other two groups.

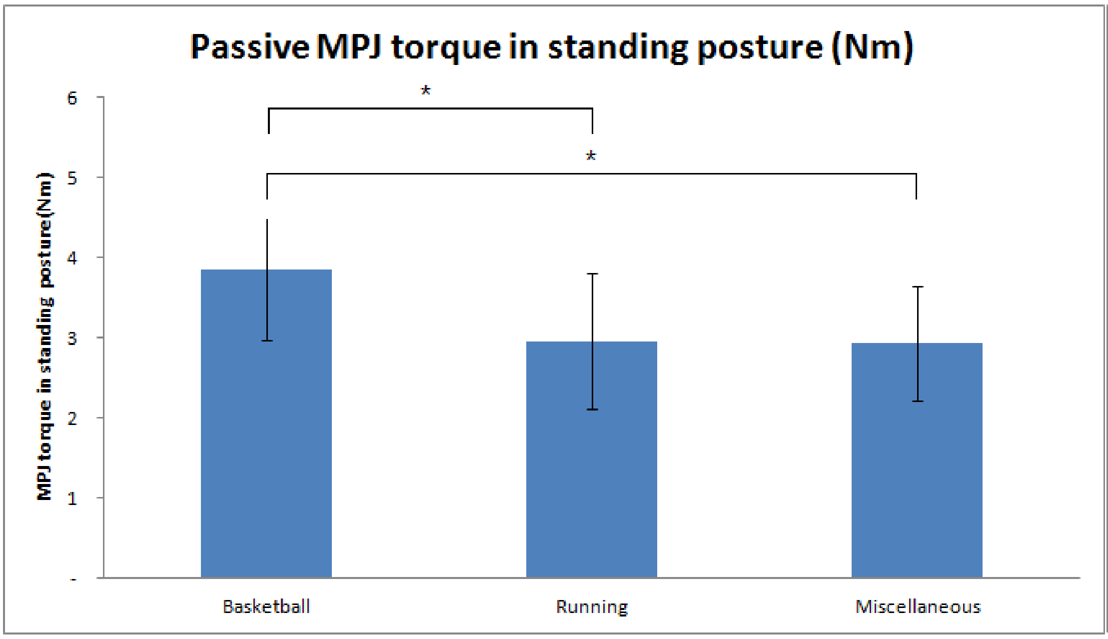


Figure 5-3 MPJ_{Stand} (Nm) among marathon runners, basketball players and miscellaneous athletes. MPJ_{stand} of basketball players was significantly higher than the other two groups.

5.6 Discussion

It is obvious that the MPJ_{sit} and MPJ_{stand} of basketball players are significantly higher than both marathon runners and miscellaneous group of athletes. It supports the first hypothesis that passive MPJ torque of basketball players is significantly higher than that of marathon runners as well as miscellaneous group of athletes. The passive MPJ torque of basketball players is approximately higher than marathon runners and other athletes by 24% and 33% respectively in both sitting and standing positions. However, T_{ankle} of basketball players was significantly higher than the miscellaneous group of athletes but not marathon runners. K_{vert} is not significantly different among sport types, which means the results do not support the second hypothesis that marathon runners have stiffer ankle joint and K_{vert} to increase the elastic energy capacity for better RE. We believe basketball game demands quick shifting, balance maintaining and jumping etc., and all these activities will strengthen toe flexors and in turn increase the MPJ_{sit} and MPJ_{stand}. While it is clear that toes play a much more important role in keeping body balance in previous study (Mueller et al., 1997). The present study suggests that body balance, a crucial factor for basketball players, is significantly influenced by toes. According to Goldmann et al., (2013), weight lifting training of toe will increase the passive stiffness of MPJ and increase jumping height and distance significantly, the current results recommend these trainings are vital for performance of basketball players.

It was reported that active ankle joint stiffness is the primary joint that dominant K_{vert} compared to the knee and the hip (Farley and Morgenroth, 1999).

However, K_{vert} does not show significant difference among the groups although T_{ankle} of basketball players was significantly higher than that of the miscellaneous group. It might be due to active joint stiffness and passive joint stiffness are largely independent and T_{ankle} is significant but only moderately correlated to K_{vert} according to the results of Chapter 4.

Applying the present dynamometer for MPJ passive joint measurements would facilitate a better understanding of sports science and also the development of treatment and rehabilitation protocols in forefoot injuries. In footwear application, optimising forefoot bending stiffness (at MPJ region) could enhance jumping, sprinting and agility performance(Stefanyshyn and Nigg, 1997). In training application, optimizing MPJ and ankle stiffness could also improve running pattern and reduce injuries (Williams et al., 2003). Monitoring MPJ stiffness could be a mean to evaluate surgical outcomes and support the design of rehabilitation protocols (intensity and duration) in forefoot related injuries (Grimston et al., 2010, Radin et al., 1978, Burr et al., 1985).

5.7 Conclusion

Basketball playing require specific body adaptation. By comparing different parameters, ankle joint stiffness of basketball players is significantly stiffer than miscellaneous sport group, and MPJ_{sit} and MPJ_{stand} of basketball players are significantly stiffer than the other two groups. Basketball players develop stronger toe plantar flexors to adapt body balance, quick shift and frequent jumping. This study suggested that strengthening both ankle and toe plantar-flexors could improve basketball players' performance.

CHAPTER 6. Passive MPJ torque, running economy, leg stiffness and vertical stiffness during sub-maximal running

6.1 Study Summary

This study examined whether passive metatarsophalangeal joints (MPJ) stiffness was associated with leg stiffness (K_{leg}) vertical stiffness (K_{vert}) and running economy (RE) during sub-maximal running. Nine male trained runners underwent passive MPJ stiffness measurements in standing and sitting positions followed by sub-maximal running on an instrumented treadmill. With the individual foot position properly aligned, the MPJ passive stiffness in both sitting (MPJ_{sit}) and standing positions (MPJ_{stand}) were measured with a computerized dynamometer. Data was collected at a running speed of 2.78 m/s, representing a stabilized level of energy expenditure. Pedar pressure insole was used to determine the contact time (t_c) and peak reaction force for the calculation of K_{leg} and K_{vert}. A respiratory gas analysis system was used to estimate the RE. Bivariate correlation test was performed to examine the correlation among MPJ stiffness, contact time, K_{leg}, K_{vert}, and RE. The results showed that MPJ_{sit} and MPJ_{stand} were inversely correlated with RE ($p=0.04$, $r=-0.68$ to -0.69), suggesting that stiffer MPJ improves RE. In addition, MPJ_{sit} was correlated positively with K_{leg} ($p<0.01$, $r=0.87$), K_{vert} ($p=0.03$, $r=0.70$) but inversely with t_c ($p=0.02$, $r=-0.76$), while MPJ_{stand} was correlated positively with the K_{vert} ($p=0.02$, $r=0.77$). These results suggested that strength of toe plantar flexors provides stability and agility in the stance phase for more effective and faster forward movement. The reduced t_c increases both K_{leg} and K_{vert}, and eventually improves RE.

6.2 Introduction

The metatarsophalangeal joints (MPJ) connect the metatarsal bones of the foot and the proximal phalanges of the toes. The MPJ play a major role in increasing the weight-bearing area (i.e., reduced plantar pressure) and enhancing body balance in walking and running (Hughes et al., 1990, Tanaka et al., 1996, Mueller et al., 1997). Clinically, trans-metatarsal amputees display inferior balance ability compared to aged-matched controls (Mueller et al., 1997). Insufficient strength of MPJ flexor muscles had also been associated with functional impairments and fall injury among the elderly (Senda et al., 1999, de Win et al., 2002, Mickle et al., 2009), while sufficient MPJ flexor strength had been associated with better shock attenuation during landing (Mann and Hagy, 1979, Senda et al., 1999). Among older adults who experienced a fall, about 70% sustain physical injuries, 25% require medical care, and 33% suffer functional decline (Stel et al., 2004). Consequently, falls among the elderly had been deemed to cause a heavy burden on the healthcare system (Alexander et al., 1992, Rizzo et al., 1996).

In sports performance research, MPJ strength can play a significant role in running performance. One EMG study indicated that toe plantar flexor muscle activation was much higher in running than in walking (Mann and Hagy, 1979) , implying that greater toe plantar flexor muscle strength is required for more vigorous activities. Moreover, it was found that the MPJ maximum plantar flexion moment was about one-fifth to one-third of the maximum ankle joint plantar flexion moment in human walking (Miyazaki and Yamamoto, 1993), suggesting that the MPJ moment is one of the contributing factors in controlling the angular momentum from the single-support to the double support phases. To this extent, the MPJ would influence running biomechanics and possibly running economy. Indeed, some studies had reported that MPJ flexor strength training would improve running performance (Hashimoto and Sakuraba, 2014). In addition to neurological responses (e.g., motor unit activation and onset frequency) and muscular changes (e.g., transition of fiber types and adenosine-triphosphate (ATP) concentration), muscle strength training exercises would influence the mechanical passive joint stiffness. For instance, passive stiffness of elbow joint improved after eccentric muscle contraction training (Pousson et al., 1990). In an animal study (i.e. rats), the optimum length of force development and passive muscle stiffness was improved with eccentric-based exercise program (Reich et al., 2000). The researchers suggested that eccentric exercises would have increased passive muscle stiffness (i.e., titin filament), which is associated with greater force generation, and thereby better athletic performance during running and jumping (Reich et al., 2000, Lindstedt et al., 2002, LaStayo et al., 2003). In a similar vein, it is highly possible that higher mechanical properties (or passive stiffness) of MPJ would partially contribute to stronger toe plantar flexors that lead to better running economy. A quantification of passive MPJ stiffness is warranted to examine the relationship between MPJ mechanical properties and running economy.

To date, the leg stiffness (K_{leg}) and vertical stiffness (K_{vert}) have been predominantly applied to describe mechanical behaviour in human locomotion (Cavagna et al., 1991, McMahon and Cheng, 1990, Seyfarth et al., 2002). K_{leg}

refers to the ratio of maximum ground reaction force ($F_{z(max)}$) to maximum change in leg length (L) during running. Previous research had suggested that the leg with higher K_{leg} can store greater elastic energy in the muscle-tendon complex during running for energy conservation (Dalleau et al., 1998). In addition, K_{vert} denotes the ratio of maximum ground reaction force ($F_{z(max)}$) to maximum vertical displacement of center of mass (CoM) during hopping (vertical jumping) or running. K_{vert} is inversely proportional to the vertical displacement of CoM, so high K_{vert} indicates that energy expended during CoM vertical movement would be reduced, thereby improving running economy (RE) (Heise and Martin, 2001). It was reported that K_{leg} and K_{vert} could be changed by gait patterns such as change of stride frequency (Farley and Gonzalez, 1996). For example, decreasing ground contact time (t_c) would increase K_{leg} (Morin et al., 2007). However, both K_{leg} and K_{vert} represent the mechanical properties of MPJ, ankle, knee and hip joints collectively, rather than the isolated properties. It thus remains uncertain whether MPJ passive stiffness would be associated with K_{leg} and K_{vert}.

Direct quantification of MPJ passive stiffness can provide essential information for clinical and sports related research, but only a few reliable MPJ stiffness measurement devices and protocols are available in the literature. Rao et al. (2011) introduced a custom-made device for measuring the MPJ passive stiffness. However, the device was manually controlled and without a mechanism for adjustment of foot alignment. The lack of appropriate foot alignment control would result in substantial measurement errors. Additionally, it is unknown whether MPJ measurement would be the best performed in sitting or standing position. Studying MPJ stiffness evaluation would provide insightful information for sports and clinical research.

In sum, the purpose of this study is to examine the correlation of MPJ stiffness, in sitting (MPJ_{sit}) and standing (MPJ_{stand}) postures, with K_{leg}, K_{vert}, gait pattern parameters (i.e. contact time (t_c), stride frequency (SF), stride length (SL)), and RE during sub-maximal running. As MPJ passive stiffness may partially contribute to the strength of toe plantar flexor muscles, which might influence gait patterns to a certain extent, it is hypothesized that MPJ passive stiffness would have significant correlations with K_{leg}, $K_{vert,}$ gait parameters, and RE.

6.3 Methods

6.3.1 Subjects

Nine male experienced runners (22 ± 3 yrs; 70 ± 8 kg; 1.78 ± 0.05 m) with regular running of 50 km per week participated in this study. They were physically active undergraduate students, with at least 6 hours exercise per week for at least 2 years so that the muscle strength reached a steady level. Only participants with a foot length of US size 9.0±0.5 for both feet were recruited. All runners were free of any lower extremity injury in the six months prior to the start of the study to avoid the joint passive stiffness was contributed by formation of fibrous tissues rather than the strength of musculatures. All participants signed an informed consent form and ethical approval was granted by human subject ethics sub-committee of The Hong Kong Polytechnic University prior to the start of the study. All the subjects finished their experiment in one single day visit.

6.3.2 MPJ torque evaluation

The MPJ_{sit} and MPJ_{stand} were first measured by the custom-made device as described in Chapter three.

6.3.3 Gait patterns, vertical and leg stiffness evaluation

The plantar pressure sensors system provided information about the t_c and $F_{z(max)}$ during sub-maximal running to determine K_{leg} and K_{vert}. The pressure data were measured using the Pedar Mobile System (Novel, GmbH, Munich, Germany), which were calibrated according to the manufacturer's guidelines. The Pedar system has been reported to have less than 2.8% of measurement error after standard calibration procedures (Putti et al., 2007). Prior to data acquisition, the instrumented insoles were inserted into both left and right test shoes. To minimize kinematic and physiological influence due to the additional mass, a data logger was supported on a tripod behind the treadmill as shown in Figure 6-1. Plantar pressures were recorded at capturing frequency of 100 Hz. Only the right leg data were used to determine K_{leg} and K_{vert} as suggested by Morin et al.,(2005). The equations for K_{leg} and K_{vert} are

$$K_{leg} = \frac{F_{z(max)}}{\Delta L} \qquad F_{z(max)} = mg\frac{\pi}{2}\left(\frac{t_c}{t_a} + 1\right)$$

$$\Delta y_{(max)} = \frac{F_{z(max)} t_c^2}{m\pi^2} + g\frac{t_c^2}{8} \qquad \Delta L = L - \sqrt{L^2 - \left(\frac{vt_c}{2}\right)^2} + \Delta y_{(max)}$$

The leg length (L) calculation using Winter's equation (leg length = 0.53 × body height) led to a mean error bias of 1.94 ± 1.51%, and the linear regression between the measured leg length and that obtained using Winter's model was significant ($R^2 = 0.89$; $p < 0.01$) (Morin et al., 2005). The treadmill running velocity (v) was 2.78m/s (i.e., 10 kilometres per hour). $F_{z(max)}$ and t_c were determined by the pressure sensors, and the gravitational constant (g) is 9.81.

The numbers of steps in trials were determined with pressure sensors. Step frequency (SF) is the average step counts in the data collection interval, and it was expressed in Hz. Step length (SL) was determined by using the relationship SL = v • SF.

Figure 6-1 A subject was running on a treadmill with pedal signal emitter hanging on a tripod behind the treadmill rather than fasting on waist for weight reduction. Wireless spirometer was used to measure RE at the same time.

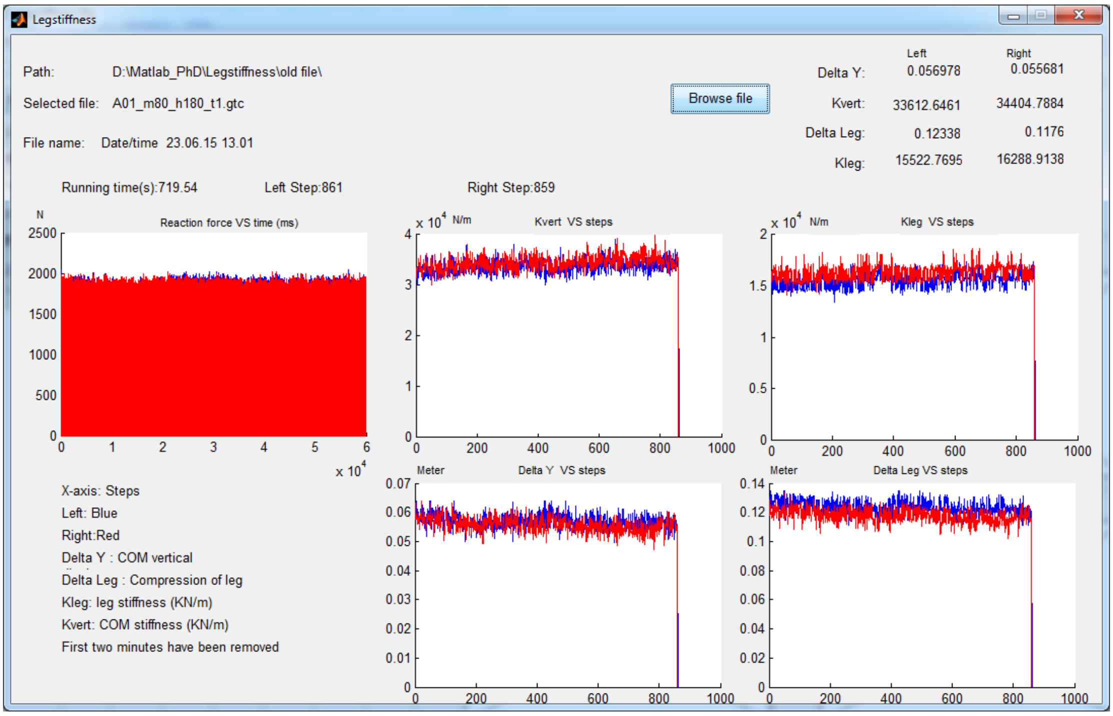


Figure 6-2 Custom made graphical user interface to show the K_{leg}, K_{vert}, ΔL and ΔY of each step during running.

6.3.4 Running economy measurement

To estimate the RE during running, the amount of oxygen consumed and the amount of carbon dioxide eliminated were measured by a wireless spirometer (Cortex Metamax 3B Spirometer, Leipzig, Germany). The participant put on a mask to record the volume of air flow, oxygen and carbon dioxide volume to determine the RE, which was expressed as ml•kg$^{-0.75}$ •min^{1}, since oxygen consumption may be better related to the -0.75 power of body mass rather than reciprocal of body mass (Bergh et al., 1991). The heart rate (HR) was recorded by a Polar Electro Sport Tester (Polar Electro, Finland) comprising a T34 heart rate transmitter attached to a strap around the participant's chest. Sampling took place at 5-second intervals.

6.3.5 Experiment procedure

After the measurement of MPJ_{sit} and MPJ_{stand}, each participant was then provided with a new pair of standard socks and a pair of test shoes (Li Ning Shoe ARBJ007) to perform a 5-minute individual stretching. The participant was then required to walk and then jog on a treadmill at speeds from 1.53 m/s to 2.50 m/s for 6 minutes as warm-up exercise. After a 10-minute rest period, the participant walked on the treadmill (Woodway PRO, Florida, US) at the speed of 1.53 m/s (1 min), and ran at 2.25 m/s (1 min), 2.78 m/s (12 min), 2.25 m/s (1 min), and finally walked at 1.53 m/s (1 min). The K_{leg}, K_{vert}, gait patterns parameters and RE data from the third to the twelve minutes (i.e., at a running speed of 2.78 m/s) were used to ensure that the participants had reached a stable running pattern in the target speed. Data were discarded if the respiratory exchange ratio went beyond 1.15 or HR over the individual theoretical value (220 – age) per minute as these are the known indicators of fatigue (Astorino et al., 2000).

6.4 Data Analysis

A custom MATLAB (Mathworks, Inc., Natwick, MA, USA) code was used to process all K_{leg}, K_{vert} and gait parameters (Figure 6-2). All torque signals were smoothened by digital Butterworth zero-lag low pass filter in 4th order with cut-off frequency of 8Hz. The MPJ stiffness was calculated by subtracting the background torque from the torque resistance and then divided by 40°. The background torque was the torque resistance when the empty pedal is swinging for the identical motion. The peak torque values of the middle ten cycles were averaged, and was defined as passive MPJ stiffness. K_{leg} and K_{vert}

of each step were determined according to the corresponding t_c and $F_{z(max)}$ and then determined from the averaged values over the total steps in the target running speed. Similarly, the gait parameters (t_c, SF and SL) were averaged. RE was computed by the manufacturer's software (Cortex Metamax 3B Spirometer, Leipzig, Germany). Pearson correlation test was performed to analyze the correlation among MPJ_{sit}, MPJ_{stand}, K_{leg}, K_{vert}, t_c, SF, SL and RE during sub-maximal running. All statistical analyses were conducted using MATLAB (Mathworks, Inc., Natwick, MA, USA) and the significance level was set at $p < 0.05$.

6.5 Results

Table 6-1 shows the individual data and the mean values of MPJ_{sit}, MPJ_{stand}, K_{leg}, K_{vert}, t_c, SF, SL and RE. Table 6-2 summarizes the Pearson correlation test between parameters, and only the parameter that significantly correlated to MPJ stiffness was presented in scatter plot in Figure 6-3.

RE was negatively correlated to both MPJ_{sit} ($p = 0.04$, $r = -0.68$) and MPJ_{stand} ($p = 0.04$, $r < -0.68$). MPJ_{sit} positively correlated to K_{leg} ($p < 0.01$, $r = 0.87$) and K_{vert} ($p = 0.03$, $r = 0.70$), and MPJ_{stand} positively correlated to K_{leg} ($p = 0.02$, $r = 0.77$). This study showed that the MPJ_{sit} was negatively correlated to t_c ($p = 0.02$, $r = -0.76$), which was inversely proportional to K_{leg} and $K_{vert.}$ However, the correlation of MPJ_{stand} to t_c ($p = 0.66$, $r = 0.17$) was not significant. In addition, RE correlated negatively to SF ($p = 0.03$, $r = -0.72$) , but positively to SL ($p = 0.05$, $r = 0.68$).

Table 6-1 The averaged MPJ passive torque, mechanical stiffness, stride frequency, contact time, relative stride length (stride length·body height^{-1}) and running economy by each subject

Subject	MPJ passive torque (Nm)		Mechanical stiffness (kNm^{-1})		Stride frequency (Hz)	Contact time (msec)	Relative stride length	Running Economy ($ml \cdot min^{-1} \cdot kg^{-0.75}$)
	MPJ_{sit}	MPJ_{stand}	K_{vert}	K_{leg}				
1	0.75	3.10	23.60	10.01	1.37	282.60	1.12	131.58
2	1.28	2.70	25.66	13.03	1.29	261.56	1.16	125.89
3	1.40	3.20	40.29	15.43	1.41	256.24	1.10	107.67
4	1.20	4.60	43.83	12.81	1.43	283.54	1.11	102.96
5	0.91	2.60	21.02	9.38	1.33	262.65	1.16	124.28
6	1.26	2.80	34.34	12.35	1.48	253.20	1.09	115.17
7	1.00	2.70	23.13	11.33	1.45	272.65	1.06	113.41
8	1.70	3.90	38.20	14.08	1.54	241.85	1.03	104.97
9	1.10	3.30	26.59	11.60	1.49	265.51	1.09	112.87
Mean	1.18	3.21	30.74	12.22	1.42	264.42	1.10	115.42

Table 6-2 The Pearson correlation test between MPJ torque, mechanical stiffness, stride frequency (SF), contact time (tc), relative stride length (SL, stride length·body height^{-1}) and running economy (RE) variables. Bolded represents the significant correlation between the test variables (P<0.05)

Parameters	*P*-value	R	Parameters	*P*-value	R
MPJ_{sit} correlate to MPJ_{stand}	0.28	0.41	K_{vert} correlate to SF	0.20	0.47
MPJ_{sit} correlate to K_{vert}	**0.03**	**0.70**	K_{vert} correlate to t_c	0.57	-0.22
MPJ_{sit} correlate to K_{leg}	**0.00**	**0.87**	K_{vert} correlate to SL	0.29	-0.40
MPJ_{sit} correlate to SF	0.19	0.48	K_{vert} correlate to RE	**0.01**	**-0.81**
MPJ_{sit} correlate to t_c	**0.02**	**-0.76**	K_{leg} correlate to SF	0.42	0.31
MPJ_{sit} correlate to SL	0.14	-0.53	K_{leg} correlate to t_c	0.17	-0.50
MPJ_{sit} correlate to RE	**0.04**	**-0.68**	K_{leg} correlate to SL	0.27	-0.41
MPJ_{stand} correlate to K_{vert}	**0.02**	**0.77**	K_{leg} correlate to RE	**0.05**	**-0.67**
MPJ_{stand} correlate to K_{leg}	0.29	0.40	SF correlate to t_c	0.36	-0.35
MPJ_{stand} correlate to SF	0.20	0.47	SF correlate to SL	**0.00**	**-0.89**
MPJ_{stand} correlate to t_c	0.66	0.17	SF correlate to RE	**0.03**	**-0.72**
MPJ_{stand} correlate to SL	0.33	-0.37	t_c correlate to SL	0.35	0.35
MPJ_{stand} correlate to RE	**0.04**	**-0.69**	t_c correlate to RE	0.48	0.27
K_{vert} correlate to K_{leg}	**0.01**	**0.78**	SL correlate to RE	**0.05**	**0.68**

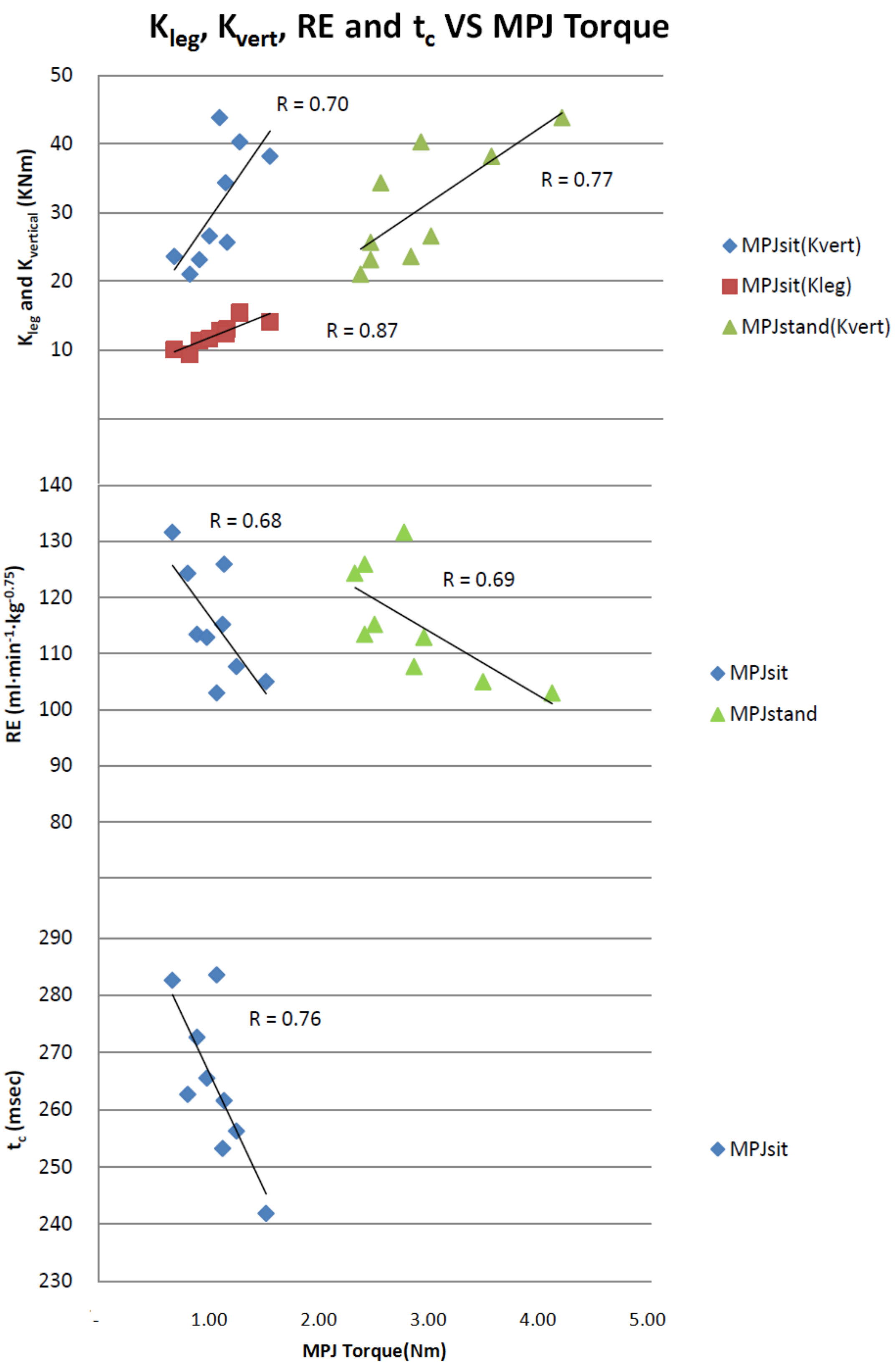


Figure 6-3 Scatter plots of leg stiffness (K_{leg}), vertical stiffness (K_{vert}), running economy (RE) and contact time (t_c) against MPJ torque in sitting and standing positions

6.6 Discussion

Apart from subjective assessment by clinicians, MPJ passive stiffness can be measured using dynamometers (Rao et al., 2011). However, current measurement techniques had failed to control both toe dorsiflexion speed and foot alignment. Without proper control of the testing conditions, it is difficult to have accurate assessment of MPJ passive stiffness. By using the customized MPJ stiffness dynamometers with controlled speed and foot alignment, this study examined if the measured MPJ_{Sit} and MPJ_{Stand} passive stiffness would be associated with K_{leg}, K_{vert}, gait variables, and RE during sub-maximal running. Firstly, the current MPJ data were comparable with those of previous studies (Rao et al., 2011), which did not control the speed of toe dorsiflexion and measured the big toe passive stiffness only. This suggests that the MPJ passive stiffness might be insensitive to speed of toe dorsiflexion or the big toe might be the major contributor for MPJ passive stiffness. In the future, studying relative passive stiffness contribution (big toe versus other toes) would possibly provide additional insights to design rehabilitation or training programs for athletes. Secondly, the MPJ passive stiffness measured in standing (MPJ_{stand}) was about 3-folds of the stiffness measured in sitting condition (MPJ_{sit}.). During standing, the plantar aponeurosis was tightened, contributing in higher the MPJ passive stiffness, which might also imply that the MPJ passive stiffness would be associated with weight bearing and ankle joint positions. Hence MPJ_{stand} would be less effective in indicating the strength of toe flexors.

The correlation results indicated that MPJ_{sit} was correlated with K_{leg}, K_{vert} and t_c, while MPJ_{Stand} was correlated only with K_{vert} (Figure 6-3). The lack of the correlation of the MPJ_{Stand} with K_{leg} can be explained by the Windlass mechanism (Hicks, 1954). During standing, the plantar aponeurosis was tightened and contributed higher MPJ stiffness, therefore MPJ_{stand} would not accurately represent the strength/passive stiffness of toes. The current results recommend that a reliable MPJ passive stiffness measurement should be performed in sitting position, rather than standing position. It should be noted that the mean K_{leg} and K_{vert} scores were determined as 12.22 kNm^{-1} and 30.74 kNm^{-1}, respectively in this study, which were 18% and 19% slower than the scores from the study by Morin et al. (2005). The discrepancies could then be due to different testing conditions and subject differences in the two studies. Specifically, Morin et al. (2005) determined K_{leg} and K_{vert} using the running speed of 3.33 m/s while our study using the running speed of 2.78 m/s. Future studies should examine if MPJ passive stiffness measured in sitting and standing would be influenced by the leg and vertical stiffness at different running speeds. For the running economy (RE), current results showed that MPJ_{sit}, MPJ_{stand}, K_{leg} and K_{vert} were negatively correlated with RE during sub-maximal running. These findings support our contention that stronger toe plantar flexors (or passive stiffness) increased the K_{leg} and K_{vert} by reducing t_c, and eventually improving RE. Although this study did not use a pre- and post-test design, it is possible that the MPJ passive stiffness was one of the important factors to reduce RE. In addition, MPJ_{sit} correlated significantly to K_{leg} and K_{vert}, which were considered to improve RE by increasing capacity of elastic energy storage in muscle-tendon unit of the leg and reducing the energy expended during vertical motion of CoM respectively. Furthermore, the

MPJ_{sit} was negatively correlated to contact time (t_c), which is inversely correlated with leg (K_{leg}) and vertical stiffness (K_{vert},). All of these correlations imply that higher MPJ_{sit} reduced t_c and thereby increased both K_{leg} and $K_{vert,}$ and eventually improved RE (i.e. decreased energy expenditure). The passive stiffness at MPJ might have been suggested to be related to running performance. Furthermore, sufficient metatarsophalangeal strength/passive stiffness had been associated with better shock attenuation during landing (Senda et al., 1999, Mann and Hagy, 1979) and the increased stiffness at forefoot would reduce energy loss at the MPJ, thus improving running performance (i.e., propulsion power and maximum speed) (Stefanyshyn and Nigg, 1997). Therefore, it is plausible that increased MPJ passive stiffness would also enhance running economy (RE). From the physiological perspective, the increased MPJ passive stiffness could result from increased strength/passive stiffness of the titin filament, which is associated with greater force generation (Reich et al., 2000, LaStayo et al., 2003, Cavagna et al., 1991)

Current results showed that higher MPJ_{sit}, MPJ_{stand}, K_{leg} and K_{vert} were negatively correlated with RE during sub-maximal running. These findings support the hypothesis that stronger toe plantarflexors increased the K_{leg} and K_{vert} by reducing t_c, and eventually improved RE.

While this study is consistent with the findings of previous studies showing that increased K_{leg} would improve RE during sub-maximal running, it might also be inferred that increased K_{leg} would help reduce to the risk of lower limb injuries.

Considering that low K_{leg} might allow excessive joint motion during running, individuals with low K_{leg} may be exposed to higher risks of soft tissue injury compared to those with high K_{leg} (Williams et al., 2003). Additionally, Granata et al., demonstrated that compared to male subjects, lower knee stiffness during hopping was found in female subjects, providing a plausible explanation why females had been reported to have a higher incidence of knee ligamentous injuries in various forms of locomotion (2002). In future studies, objective measurement of MPJ passive stiffness (e.g., MPJ_{sit}) with a dynamometer may be used as a simple and effective assessment (i.e., no motion analysis or plantar force measurement needed) to estimate leg stiffness (K_{leg}) when implementing toe strengthening programs to minimize the risks of injuries and enhance performance in running.

The dynamometer used in this current study determined the correlation between passive MPJ stiffness and lower limb kinematics data. The computerized control MPJ passive stiffness measurement, along with appropriate foot alignment would allow researchers and clinicians to systematically measure the MPJ stiffness with high reliability for both sports and clinical research to support enhanced performance. For example, optimized MPJ stiffness facilitated by footwear design could enhance jumping, sprinting and agility performance (J. Stefanyshyn and Nigg, 1998). Optimizing MPJ stiffness by training, could also improve running pattern and reduce injuries (Williams et al., 2003). Moreover, monitoring MPJ stiffness could be a means to evaluate surgical outcomes and support the design of rehabilitation protocols (intensity and duration) in forefoot related injuries (Grimston et al., 2010, Radin et al., 1978, Burr et al., 1985). It should be noted that only nine runners participated in this study. In the future, a bigger sample size would allow researchers to examine the interplay between genders (Miller et al., 1993, Krauss et al., 2008), age, and foot morphology. Further investigations in this area will facilitate a better understanding of sports science and also the development of treatment and rehabilitation protocols in forefoot injuries.

6.7 Conclusion

This study showed that running economy, and leg and vertical stiffness during sub-maximal running was correlated with MPJ passive stiffness measured in sitting position for most parameters, except for the correlation of the passive stiffness with leg stiffness. Better running economy may be explained by the reduction of contact time and/or increase of MPJ stiffness. It is therefore suggested that running economy during submaximal running would be improved by increased toe flexor strength.

CHAPTER 7. Conclusion and Future Research

7.1 Conclusion

7.1.1 Summary of key findings

One of the prime findings was that the ICC of the custom made dynamometer for measurement was over 0.9, which means that the device is systematic and reliable for the measurements and can be used for multiple scientific and clinical purposes. To the best of my knowledge, this is the first computer controlled dynamometer that can measure MPJ and ankle joint torque. This can be used to show how toe muscles training programs can affect toe passive stiffness, and how does toe passive stiffness relate to physical performance in terms of agility and running economy, etc.

In addition, the passive MPJ and the ankle joint torques were significantly correlated to K_{vert} during hopping ($p<0.005$, $R=0.29$ to 0.43). The positive correlation indicated that the high passive MPJ and ankle joint stiffness resisted the CoM displacement during hopping. However, the weak correlation also showed that there were other determinant factors that affected the K_{vert}. Besides, the T_{ankle} and MPJ_{sit} and MPJ_{stand} were also significantly correlated ($p<0.001$, $R=0.413$ to 0.718). The correlation were ranged from weak to moderate only. During walking and running, both the ankle joint and MPJ plantar-flex in the push off phase to provide forward and upward propulsion. It is reasonable to expect that the strength of the both joints are highly correlated. The present results suggested that the MPJ and ankle joint may provide different functions in different circumstances.

The T_{ankle}, MPJ_{sit}, MPJ_{stand} and K_{vert} acquired from the basketball players, marathon runners and miscellaneous type of athletes were also compared. It was found that MPJ_{sit} and MPJ_{stand} of basketball players were significantly higher than other groups, which led to conclude that basketball players need to be better at jumping, keeping balance and changing directions. These actions require stronger toe plantar flexors and basketball players should develop these characteristic for adaptation. It was reported that joint passive stiffness could be increased by eccentric exercise (Porter et al., 2002), and strengthening toe flexor muscles can enhance physical performance in term of jumping height and jumping distance (Goldmann et al., 2013). The current study agreed previous findings and suggested that toe flexor training would be introduced to enhance physical performance especially for basketball players.

The MPJ_{sit}, MPJ_{stand}, K_{leg} and K_{vert} and RE during sub-maximal running were also examined. It was found that MPJ_{sit} and MPJ_{stand} were correlated negatively with RE in significant level ($p=0.04$, $r=-0.68$ to -0.69), suggesting that stiffer MPJ improves RE. The improvement of RE can be explained by the reduction of contact time with increase of toe plantar-flexor muscles strength, and hence increase the K_{leg} and K_{vert} significantly.

7.1.2 Project significances and values

A new computer controlled dynamometer which can be used to measure both MPJ and ankle joint stiffness was developed. It allows researchers and clinicians systematically measure the stiffness of both joints in controlled motion. In sport science application, MPJ stiffness can be acquired to estimate the strength of toes in order to evaluate the body agility (Mann and Hagy, 1979, Mickle et al., 2009). Since the MPJ stiffness is also significantly correlated to K_{leg} in sub-maximal running, it would be associated with sport injuries since K_{leg} was related to lower limb injury (Grimston et al., 2010, Radin et al., 1978, Burr et al., 1985). Adjusting MPJ stiffness by appropriate training may alter running form and hence reduce injuries. In clinical application, MPJ stiffness and range of motion can be the evaluation index of healing process of after toe-related surgery or treatment.

7.2 Limitations of Studies

7.2.1 Gender of subjects

Only male subjects were recruited in this study., therefore the influence of gender was not demonstrated (Miller et al., 1993). Muscle size and fiber characters are different between male and female (Miller et al., 1993), and may affect the outcome of measuring passive stiffness of the ankle and MPJ. Furthermore, the running biomechanics between male and female may be different (Decker et al., 2003, Ferber et al., 2003, Lephart et al., 2002). This may affect the results of RE experiment.

7.2.2 Uneven distribution of subject pool

In this study, only healthy and active subjects with normal feet were recruited. Subjects with foot deformities, such as hallux valgus, were excluded. Moreover, only experienced runners were recruited for the running economy experiment. Thus the experimental results may not represent the recreational runners.

7.2.3 Treadmill running

Treadmill running is not equivalent to ground running. Wank et al. (1998) reported that kinematics and electromyography of treadmill running and ground running were different. It may limit the results of this study to apply to field test.

7.2.4 Improvement of the dynamometer

The current dynamometer adjusts the height of footplate by inserting different thickness of plastic plates for alignment. It can be a time consuming process. An automatic height adjustment mechanism will enhance the measuring process.

7.3 Directions of Future Research

Muscle sizes and fiber characters are different between male and female, these may lead to differences in MPJ and ankle joint stiffness between gender. It is suggested to recruit female subjects for comparison.

Strain energy can be defined as the product of torque resistance and angular displacement. It is stored and released in the joints during flexion and extension. The dynamometer can be used to measure the strain energy of MPJ and ankle joint, since it provides passive MPJ and ankle joint flexion and extension and measures the torque resistance at the same time. Strain energy storage capacity of MPJ and ankle joints may be important in running economy.

The dynamometer can measure the stiffness of hinge joint of ankle foot orthosis with or without the foot putting on, and clinician can select orthotic hinge joint with suitable rigidity.

For the study in Chapter six, although the results support the hypothesis that running economy is significantly and negatively correlated to MPJ torque, leg stiffness and vertical stiffness, only nine subjects participated in the experiment. It is suggested to recruit more subjects in order to increase the confidence level.

Implement clinical evaluation for toe deformity. There are different surgical approaches for same type of toe deformity, but there is no specified tool to quantified the surgical treatment. The present device can quantitatively evaluate the outcome in term of range of motion, stiffness and maximum force output of the toes.

The passive extension stiffness of the toes partially comes from the tension of plantar aponeurosis and the extrinsic muscles of the toes. These structures also maintain the normal shape of foot arch. Foot deformity such as flatfoot and high arch foot may affect the passive MPJ stiffness. It is worth to investigate the relationships of toe stiffness and the arch height index.

Apart from passive torque measurement, the current device can measure maximum isometric voluntary contraction of toes and ankle joint by locking the cradle at particular inclined angle. The active and passive MPJ and ankle joint moments can be correlated to understand their relationships.

Low leg stiffness might allow excessive joint motion and cause lower limb soft-tissue injuries. Since the results of this experiment indicated that MPJ stiffness was significantly correlated with leg stiffness, it was suggested that leg stiffness may be manipulated by toe strengthening programs and hence reduce the risk of lower limb soft tissue injuries. Further studies should be introduced in this direction.

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APPENDIX A Consent form sample in Chapter 3 to 5

LI-NING

THE HONG KONG POLYTECHNIC UNIVERSITY 香港理工大學

Research Design by Li Ning Sports Science Research Center and The Hong Kong Polytechnic University

The Effect of Anthropometry and Foot Dimensions on Passive MPJ Stiffness and Active MPJ Moment of Male Athletes

You are invited to participate in a research named above from the Department of Health Technology and Informatics, The Hong Kong Polytechnic University. You will understand the research protocol and sign the Informed Consent before starting the research. Please read the page 2 for the experiment procedures.

All information revealing your identity will be kept confidential, only coded for research purpose. You have the right to withdraw from this research any time without penalty. If you have any concerns about the conduct of this research study, please do not hesitate to contact Ms Kath Lui (Tel: 852-27667933, Fax: 852-23557651), Secretary of the Human Subjects Ethics Sub-Committee of The Hong Kong Polytechnic University in person or in writing (c/o Human Resources Office of the University).If you would like more information about the research, you are welcomed to contact Dr. Aaron Leung at 27667676. Thank you very much again for joining this research.

1

Research Design by Li Ning Sports Science Research Center
and The Hong Kong Polytechnic University

The Effect of Anthropometry and Foot Dimensions on Passive MPJ Stiffness and Active MPJ Moment of Male Athletes

Experiment Procedures

1. Body height , weight and foot size were recorded.
2. Both feet of subject were scanned by an optical foot scanner.
3. Hold the hands on their waist and hopped for 17 seconds in preferred height with both, right and left legs on a force plate respectively. Subject had a rest between each trial.
4. Subject walked on a treadmill for 5 minutes in 1.5km/s as a preconditioning excise.
5. In sitting position, a foot was fasted on foot platform with straps. The foot was passively moved from 20° plantar flexion to 20° dorsi-flexion in angular speed of 40°/s for 20 cycles. Remove the foot and repeat the procedure with another foot.
6. In sitting position, a foot was fasted on the foot platform with straps on another device. The toes were passively moved from 0° to 40° in angular speed of 40°/s for 20 cycles. Repeat the procedure in standing position.
7. Repeat the procedure 6 with another foot.
8. End of experiment.

Research Design by Li Ning Sports Science Research Center and The Hong Kong Polytechnic University

CONSENT FORM

Project Title: The Effect of Anthropometry and Foot Dimensions on Passive MPJ Stiffness and Active MPJ Moment of Male Athletes

Name of Researcher: Man Hok Sum

	Tick to Confirm
• I confirm that I have read and understand the information sheet for the above study.	☐
• I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.	☐
• I know that all information revealing my identity will be kept confidential	☐
• I understand that my participation is voluntary and that I am free to withdraw from this research any time without penalty.	☐
• I agree to participate this research study.	☐

Name of participant	Date	Signature
Researcher	Date	Signature

Research Design by Li Ning Sports Science Research Center
and The Hong Kong Polytechnic University

男性运动员的足部尺寸和形状对足部关节柔软度和力量与垂直跳跃能力的影响

欢迎您参与由香港理工大学医疗科技及资讯学系的以上研究。

在开始任何测试前，您将了解研究内容和签署同意书。研究步骤请参阅第 2 页。

此计划中所有能辨识您的身份的资料都会受到保密，只供研究用途。您有权利在开始测试期间退出研究，而不会受到任何惩处。

如果您对这研究有任何意见或投诉，您可以联络本校的道德委员会的吕小姐 (Tel: 852-27667933， Fax: 852-23557651)，或研究负责人梁锦伦博士（电话联络:852-27667676）。

再次感谢您参与是次研究。

Research Design by Li Ning Sports Science Research Center
and The Hong Kong Polytechnic University

男性运动员的足部尺寸和形状对足部关节柔软度和力量与

垂直跳跃能力的影响

<u>实验步骤</u>

1. 记录参加者的身高、体重和脚的大小。
2. 在双脚放置标记并以用光学扫描器扫描脚部，记录脚的形状。然后把反射标记移除。
3. 在力台上随着节拍器的频率用(1)左腿、(2)右腿和(3)双腿垂直跳 17 秒。在各试验之间有休息时间。
4. 在跑步机上以 1.5m/s 的速度步行 5 分钟作热身体运动。
5. 坐在椅子上，用带子把脚固定在仪器上，脚部从 20°向上屈曲转动至 20°向下屈曲重复 20 次。把脚从脚台移除，以另一只脚重复步骤。
6. 坐在椅子上，用带子把一脚掌固定在另一台仪器上，脚趾向上屈曲 40°重复 20 次。然后以站立姿势重复此步骤。
7. 把脚从仪器移除，以另一只脚重复步骤 6。
8. 实验结束。

Research Design by Li Ning Sports Science Research Center
and The Hong Kong Polytechnic University

同意书

项目名称：男性运动员的足部尺寸和形状对足部关节柔软度和力量与垂直跳跃能力的影响

研究员名称: 文学森

	确认
• 我确认我已阅读并了解上述研究的资料	□
• 我有机会考虑和提问，并得到满意的答案。	□
• 我知道所我的个人资料都会受到保密	□
• 我明白我是自愿参与的，我可以在任何时间退出本研究而不受惩罚。	□
• 我同意参与此项研究	□

参加者姓名　　日期　　签署

研究员姓名　　日期　　签署

3

APPENDIX B Ethical approval by the Human Ethics Sub-committee of The Hong Kong Polytechnic University

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

To Leung Kam Lun (Interdisciplinary Division of Biomedical Engineering)

From ZHANG Ming, Chair, Departmental Research Committee

Email htmzhang@ Date 12-Nov-2012

Application for Ethical Review for Teaching/Research Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following project for a period from 12-Nov-2012 to 30-Jan-2015:

Project Title:	The Effect of Anthropometry and Foot Dimensions on Metatarsophalangeal Joint (MPJ)Stiffness
Department:	Interdisciplinary Division of Biomedical Engineering
Principal Investigator:	Leung Kam Lun

Please note that you will be held responsible for the ethical approval granted for the project and the ethical conduct of the personnel involved in the project. In the case of the Co-PI, if any, has also obtained ethical approval for the project, the Co-PI will also assume the responsibility in respect of the ethical approval (in relation to the areas of expertise of respective Co-PI in accordance with the stipulations given by the approving authority).

You are responsible for informing the Departmental Research Committee in advance of any changes in the proposal or procedures which may affect the validity of this ethical approval.

You will receive separate email notification should you be required to obtain fresh approval.

ZHANG Ming

Chair

Departmental Research Committee

证书号第1817044号

发明专利证书

发明名称：用于关节扭矩测量及训练系统

发明人：文学森;梁諦伦;张德文;托斯滕·斯特辛

专利号：ZL 2014 1 0299533.4

专利申请日：2014年06月26日

专利权人：李宁体育（上海）有限公司

授权公告日：2015年10月14日

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局长
申长雨 [redacted]

中华人民共和国国家知识产权局
2015年10月14日

第1页（共1页）

APPENDIX D Raw data

Background information					
Subject	Age(yrs)	Height(cm)	Mass(kg)	Left foot size (US)	Right foot size(US)
1	21	174	67	7.5	7.5
2	24	181	82	9	9.5
3	25	174	65	10	9.5
4	24	174	64	8.5	8
5	23	171.5	75.2	8	8
6	24	173.5	62.9	9	8.5
7	30	180	79.4	10	10.5
8	19	176	77	9.5	9.5
9	19	179.5	65	9.5	9
10	26	167	57.8	7	6.5
11	24	189	85.6	12	12
12	24	168	70.4	7.5	7
13	24	175	63.5	9	9
14	23	175	83.6	9	8.5
15	25	184	81.6	10.5	10
16	22	178	88.7	9	8.5
17	22	179.5	62.4	9.5	9
18	20	174	67.6	8.5	8
19	22	181.5	82.2	9.5	9
20	24	176	74	10	9
21	21	176	72.8	9	9
22	23	179.5	76.6	10.5	10
23	20	176.5	77.1	9	8.5
24	21	183	80	10.5	10.5
25	20	172	62.4	7.5	7
26	19	178.5	60.8	8	7.5
27	26	174.5	77.3	11	10.5
28	24	184.5	75.7	8.5	9
29	23	174	69.5	8	8
30	24	178	73.2	8.5	8
31	23	168	64.7	8.5	8
32	23	168	64.7	8.5	8
33	21	179.5	59.5	9	9.5
34	22	181.9	103	11.5	11
35	22	184.5	82.5	10.5	10
36	23	195	103.5	12	11.5
37	21	180.5	72.5	10.5	10.5

Background information					
Subject	Age(yrs)	Height(cm)	Mass(kg)	Left foot size (US)	Right foot size(US)
38	21	184	100.7	11.5	12
39	21	179.5	84.3	10	9.5
40	20	182.5	74.7	12.5	12.5
41	20	193.5	93.5	12.5	12
42	26	167	59.4	8	8
43	23	182.5	76.5	11	10.5
44	21	183	66.2	10	9
45	22	187	117.6	12	11.5
46	20	187	81	12	11.5
47	22	188	85.5	10.5	9.5
48	20	191.5	83.5	12	11.5
49	27	164	70	5.5	5.5
50	25	165	62	6.5	6.5
51	20	167	66	8	7.5
52	21	182	74	10	10
53	22	184	75	10	10
54	22	185	70	10	10
55	22	163	63	6.5	7
56	22	164.5	68	6.5	6
57	23	178	77.2	8	8
58	19	170	60	6	5.5
59	21	170	62	7.5	7
60	25	197	88	14.5	13.5
61	19	195	94	10.5	10
62	23	187	72	9.5	9.5
63	22	174	58	7.5	7
64	22	165	59	8.5	8
65	21	163	58	6.5	7
66	19	168	52	6.5	6
67	22	165	54	7.5	7
68	20	173	70	8.5	8
69	21	171	82	7.5	7
70	23	165	59	6.5	6
71	24	171	77	7.5	7
72	20	170	60	6.5	6.5
73	21	170	63	8.5	8
74	24	166	56	6.5	6
75	22	167	68	7.5	7.5
76	23	192	110	11	10.5

Background information					
Subject	Age(yrs)	Height(cm)	Mass(kg)	Left foot size (US)	Right foot size(US)
77	21	185	78	11.5	11
78	23	193	91.2	13.5	12.5
79	20	181.5	72.4	12	11.5
80	23	164	80.7	7	6.5
81	25	171.5	68	7	7.5
82	25	174	64	6.5	6.5
83	28	170	64	8.5	8.5
84	25	168.5	68	8	7.5
85	27	185.5	78.4	11.5	12
86	20	183.5	71.6	10.5	10.5
87	25	178.5	72.6	10	9.5
88	21	184.5	92.1	12	11.5
89	32	179	88	10	9.5
90	19	179	67	9	8.5
91	19	181	66	9.5	9.5
92	20	178	74	9	9
93	19	170	65	8	8
94	19	176	80	8	8.5
95	25	176	76.6	9.5	9.5
96	23	175.5	73	8	8
97	23	184	81	10	10
98	24	167	66	7.5	8
99	24	169	62	6.5	6.5
100	25	168	66	7.5	7

	Exercise type & frequency		
Subject	Regular exercise type	Exercise experience(yrs)	Exercise exposure (km/wk)
1	Tennis	3	6
2	Running	5	15
3	Basket Ball, Tennis, R	10	2
4	Running	3	30
5	Other	3	15
6	Table tennis, Badminton	6	3
7	Badminton	5	1
8	Basket Ball	6	6
9	Basket Ball	6	4
10	Table tennis	5	2
11	Football	9	4
12	Football	10	2
13	Basket Ball	8	2
14	Other	5	25
15	Basket Ball	12	2
16	Running	3	5
17	Running	2	5
18	Basket Ball	6	3
19	Basket Ball	6	4
20	Running	6	4
21	Basket Ball	2	1
22	Basket Ball	4	3
23	Basket Ball	7	4
24	Basket Ball	4	3
25	Tennis	1	5
26	Tennis	1	5
27	Tennis	4	4
28	Basketball	1	5
29	Basketball	5	2
30	Basketball	6	4
31	Basketball	6	3
32	Basketball	6	3
33	Table Tennis	3	2
34	Basketball	6	2
35	Basketball	8	4
36	Basketball	8	3.5
37	Basketball	10	2
38	Football	8	1.5
39	Basketball	10	2.5

	Exercise type & frequency		
Subject	Regular exercise type	Exercise experience(yrs)	Exercise exposure (km/wk)
40	Football	8	3
41	Basketball	8	3.5
42	Running	18	50
43	Basketball	6	3.5
44	Basketball	6	2.5
45	Other	10	6
46	Basketball	4	2
47	Basketball	5	3
48	Basketball	10	5
49	Basketball	11	5.5
50	Tennis	6	3.5
51	Basketball	8	3.5
52	Soccer	7	2
53	Basketball	4	3.5
54	Basketball	8	3
55	Table Tennis	3	3
56	Table Tennis	4	4
57	Volleyball	3	3
58	Running	9	4
59	Running	5	4
60	Basketball	10	2.5
61	Basketball	3	1
62	Badminton	2	14
63	Badminton	3	2
64	Other	2	3
65	Other	2	3
66	Running	5	4
67	Running	3	6
68	Basketball	4	2
69	Basketball	5	2
70	Running	3	3
71	Badminton	2	2.5
72	Basketball	3	2
73	Other	2	3
74	Running	5	5
75	Other	2	2
76	Running	13	3
77	Basketball	9	4
78	Running	3	4

	Exercise type & frequency		
Subject	Regular exercise type	Exercise experience(yrs)	Exercise exposure (km/wk)
79	Basketball	7	4
80	Other	12	5
81	Basketball	3	3
82	Running	12	6
83	Soccer	10	6
84	Other	2	7
85	Basketball	7	3
86	Basketball	6	4
87	Running	12	5
88	Basketball	7	6
89	Running	10	25
90	Running	6	8
91	Running	3	8
92	Gym	4	10
93	Running	3	8
94	Gym	6	15
95	Gym	10	40
96	Gym	7	10
97	Gym	8	40
98	Running	8	10
99	Gym	6	8
100	Running	4	10

	Passive joint stiffness (Nm)					
Subject	Ankle L	Ankle R	MPJ L sit	MPJ L stand	MPJ R sit	MPJ R stand
1	6.13	7.97	0.83	1.96	0.99	2.00
2	11.41	7.26	2.14	3.88	1.77	3.38
3	7.80	7.28	1.95	5.77	1.93	6.18
4	6.16	5.53	0.89	2.07	1.23	2.20
5	9.62	7.65	1.92	3.75	1.83	5.12
6	7.86	8.70	1.45	3.39	1.40	3.38
7	5.48	7.72	1.22	2.79	1.49	3.16
8	7.53	6.44	1.71	3.74	1.52	3.93
9	8.69	7.60	1.71	3.87	1.67	4.65
10	5.64	5.38	1.15	2.74	1.18	2.45
11	10.20	9.59	2.27	3.60	1.23	2.40
12	6.54	8.33	1.29	3.77	1.42	3.98
13	6.08	6.18	1.42	4.88	1.39	2.99
14	5.78	6.24	1.25	3.95	1.15	3.14
15	8.76	10.59	1.46	5.24	1.46	4.10
16	9.05	9.79	1.75	5.73	2.77	5.35
17	7.68	5.64	1.27	2.47	1.41	2.68
18	10.12	11.05	1.68	5.05	2.03	6.46
19	11.76	10.72	2.76	4.80	2.07	4.45
20	7.34	6.29	1.08	3.83	0.98	3.14
21	12.52	10.89	1.44	2.86	1.67	4.16
22	8.12	5.79	1.76	4.34	1.50	2.57
23	13.28	13.44	3.20	5.33	2.55	5.35
24	9.06	10.67	2.47	6.21	1.78	5.79
25	5.41	5.30	1.00	2.15	0.72	2.19
26	6.55	8.93	1.28	3.78	1.05	5.18
27	11.70	13.38	2.36	4.95	2.59	5.76
28	9.88	7.52	1.41	2.61	1.03	2.51
29	6.10	6.85	1.08	3.01	1.10	3.07
30	6.41	6.79	1.77	3.61	1.72	2.85
31	8.45	9.52	1.68	3.37	1.92	3.54
32	6.86	7.57	1.17	3.04	1.40	3.66
33	8.06	9.94	1.59	2.40	1.87	3.19
34	7.76	8.08	1.62	4.42	1.93	5.82
35	9.64	9.46	1.80	4.00	1.66	3.15
36	12.16	12.52	1.71	4.51	1.48	3.59
37	6.26	7.17	1.29	4.43	1.13	3.37
38	6.21	8.06	1.13	2.28	1.29	1.97
39	10.31	7.80	1.59	5.27	1.57	4.91

	Passive joint stiffness (Nm)					
Subject	Ankle L	Ankle R	MPJ L sit	MPJ L stand	MPJ R sit	MPJ R stand
40	8.91	7.50	1.28	4.28	1.64	4.09
41	5.79	5.22	0.99	4.32	1.06	4.33
42	6.11	5.50	1.06	3.12	1.01	3.67
43	12.07	11.89	2.00	5.43	2.09	5.34
44	7.59	6.09	1.19	3.99	1.12	3.00
45	14.12	14.63	1.89	2.83	2.03	3.21
46	8.70	7.84	1.28	3.41	1.62	4.31
47	11.83	10.11	1.17	1.78	0.96	3.30
48	10.80	7.88	1.66	3.34	1.58	3.33
49	5.26	5.59	1.23	2.12	1.37	1.82
50	7.50	9.63	0.86	2.55	0.99	2.83
51	5.80	5.40	1.02	3.05	1.11	2.96
52	6.03	5.19	1.53	2.76	1.37	3.79
53	12.32	10.16	1.86	4.00	1.80	4.66
54	13.16	11.35	2.00	3.63	2.20	4.18
55	10.21	8.30	1.02	2.87	1.19	2.35
56	6.00	5.19	1.32	2.81	0.97	2.78
57	9.47	7.12	1.24	2.69	0.98	2.61
58	6.98	6.72	0.86	2.12	0.82	2.25
59	6.92	7.49	1.52	3.42	2.19	4.52
60	7.05	6.26	1.18	2.52	0.91	2.33
61	5.49	5.38	0.93	2.73	1.08	3.10
62	6.27	5.91	1.21	2.77	1.13	4.19
63	6.37	6.61	1.53	3.05	1.34	2.68
64	5.46	5.03	1.07	2.70	0.85	2.10
65	6.42	5.89	1.36	3.04	0.95	2.98
66	5.24	4.87	0.66	1.31	0.76	1.65
67	3.27	3.37	0.79	1.99	0.54	1.72
68	6.16	5.22	1.22	3.34	1.47	4.48
69	9.20	8.81	1.26	3.52	1.26	2.67
70	7.24	7.52	1.05	2.45	1.19	2.66
71	6.66	7.21	0.83	1.89	1.05	2.36
72	13.68	11.67	1.55	4.50	1.50	4.22
73	5.57	5.32	1.03	1.67	0.83	1.79
74	9.46	10.42	1.06	1.87	1.12	2.36
75	8.77	9.31	1.35	3.03	1.33	2.61
76	9.86	10.36	1.64	3.59	1.60	5.18
77	6.90	6.40	1.48	4.20	1.38	4.43
78	11.25	11.28	1.77	4.63	1.66	4.78

	Passive joint stiffness (Nm)					
Subject	Ankle L	Ankle R	MPJ L sit	MPJ L stand	MPJ R sit	MPJ R stand
79	12.04	10.92	1.92	4.11	1.98	4.60
80	7.95	8.04	1.29	1.90	1.75	2.67
81	7.93	7.89	1.34	2.60	1.47	2.59
82	4.96	4.47	1.11	3.29	0.99	3.19
83	7.11	9.13	1.57	3.98	1.44	2.89
84	6.19	5.76	1.24	3.37	1.10	3.36
85	13.15	15.73	2.11	4.60	2.24	7.08
86	6.19	5.80	0.91	2.87	1.09	2.43
87	8.86	10.21	1.16	2.89	1.13	3.97
88	10.26	12.39	2.03	5.84	2.33	5.11
89	10.01	11.72	1.76	5.14	1.81	5.13
90	9.42	9.48	1.20	2.00	0.79	1.65
91	9.56	8.79	0.84	1.45	0.95	1.83
92	7.38	8.12	1.44	5.20	1.54	5.09
93	5.54	7.15	0.77	3.19	0.89	2.24
94	14.81	13.29	1.13	3.84	0.91	3.11
95	13.34	10.70	1.76	4.17	1.78	2.51
96	10.17	9.55	1.11	2.47	1.17	2.90
97	13.93	13.31	2.23	6.18	1.94	6.60
98	9.20	8.79	1.28	2.33	1.45	2.94
99	8.20	7.87	1.03	2.67	1.33	3.26
100	6.29	8.59	1.42	3.06	1.17	3.22

	Hopping (kN/m)		
Subject	Both	Left	Right
1	12.20	9.52	9.60
2	17.00	12.45	11.82
3	10.36	8.35	7.61
4	18.35	12.64	11.90
5	16.61	11.22	10.98
6	11.07	8.71	8.13
7	16.82	11.54	13.06
8	15.40	11.22	11.65
9	17.18	10.04	10.45
10	14.61	9.54	10.27
11	19.29	10.11	11.73
12	19.27	11.00	10.42
13	18.06	8.40	8.87
14	14.55	9.69	9.74
15	15.69	8.87	9.82
16	18.94	12.69	12.36
17	17.78	8.28	8.29
18	18.36	11.56	11.01
19	16.83	13.79	14.26
20	15.72	11.13	10.90
21	15.28	12.29	12.56
22	13.26	10.47	10.91
23	19.71	11.23	9.55
24	13.21	9.93	9.93
25	15.95	9.03	8.47
26	14.69	8.98	9.98
27	10.56	8.12	8.66
28	15.31	10.79	9.17
29	15.43	11.53	10.47
30	16.81	10.56	9.86
31	12.35	8.35	9.19
32	9.71	7.97	7.75
33	14.14	9.30	9.81
34	15.63	13.91	13.32
35	14.25	10.99	10.69
36	15.81	12.15	11.86
37	13.19	9.79	9.71
38	16.15	11.21	11.51
39	11.04	8.17	7.96

	Hopping (kN/m)		
Subject	Both	Left	Right
40	16.08	10.69	10.86
41	16.30	11.25	10.74
42	11.79	7.92	7.94
43	12.72	9.75	9.01
44	10.50	8.07	8.97
45	20.41	13.12	13.77
46	10.04	10.36	9.14
47	11.88	12.94	12.30
48	23.25	11.75	11.92
49	9.50	10.31	7.17
50	8.29	7.21	8.13
51	17.48	8.37	7.77
52	14.55	9.23	8.44
53	17.94	10.62	9.82
54	15.06	8.88	8.98
55	9.57	7.90	8.25
56	7.84	7.75	7.50
57	13.32	9.68	9.14
58	10.37	8.09	7.68
59	14.18	10.64	10.56
60	14.89	11.54	10.38
61	15.57	11.14	11.78
62	15.84	9.83	10.70
63	14.78	9.10	9.07
64	9.88	7.67	7.03
65	12.71	7.98	7.97
66	10.15	6.81	7.19
67	9.14	6.77	6.86
68	10.51	7.78	8.65
69	13.68	10.00	9.84
70	11.84	7.58	8.28
71	15.61	12.60	12.06
72	18.39	11.70	11.89
73	12.00	8.46	9.51
74	10.88	8.35	8.20
75	14.59	10.54	10.47
76	17.05	12.07	13.38
77	16.03	12.63	12.76
78	19.01	12.28	12.52

	Hopping (kN/m)		
Subject	Both	Left	Right
79	12.44	10.75	10.77
80	21.17	11.75	12.74
81	13.55	10.24	10.82
82	10.21	7.93	8.36
83	12.82	11.19	12.02
84	16.20	10.26	9.65
85	24.54	12.76	14.60
86	13.25	9.46	10.82
87	13.33	10.49	10.16
88	14.39	12.86	9.55
89	16.58	12.05	12.81

	Hopping (kN/m)		
Subject	Both	Left	Right
90	14.46	9.90	8.49
91	15.20	9.69	10.03
92	17.55	12.45	11.14
93	12.42	9.00	8.10
94	14.64	12.35	10.61
95	16.89	11.53	10.88
96	13.28	9.86	9.96
97	15.74	10.49	11.07
98	14.38	7.74	7.66
99	16.53	8.08	8.88
100	11.05	9.25	8.46

REFERENCES

Abdelkrim, N. B., El Fazaa, S. & El Ati, J. 2007. Time–motion analysis and physiological data of elite under-19-year-old basketball players during competition. *British journal of sports medicine,* 41**,** 69-75.

Akhbari, B., Takamjani, I. E., Salavati, M. & Sanjari, M. A. 2007. A 4-week biodex stability exercise program improved ankle musculature onset, peak latency and balance measures in functionally unstable ankles. *Physical Therapy in Sport,* 8**,** 117-129.

Alexander, B. H., Rivara, F. P. & Wolf, M. E. 1992. The cost and frequency of hospitalization for fall-related injuries in older adults. *American journal of public health,* 82**,** 1020-1023.

Alexander, R. 1987. The spring in the arch of the human foot. *Nature,* 325**,** 147-149.

Alexander, R., Ker, R., Bennet, M., Bibby, S. & Kester, R. 1987. The spring in the arch of the human foot. *Nature,* 325**,** 147-149.

Alexander, R. M. 1991. Energy-saving mechanisms in walking and running. *Journal of Experimental Biology,* 160**,** 55-69.

Amin, D. J. & Herrington, L. C. 2014. The relationship between ankle joint physiological characteristics and balance control during unilateral stance. *Gait & posture,* 39**,** 718-722.

Araújo, V. L., Carvalhais, V. O., Souza, T. R., Ocarino, J. M., Gonçalves, G. G. & Fonseca, S. T. 2011. Validity and reliability of clinical tests for assessing passive ankle stiffness. *Brazilian Journal of Physical Therapy,* 15**,** 166-173.

Arampatzis, A., Brüggemann, G.-P. & Metzler, V. 1999. The effect of speed on leg stiffness and joint kinetics in human running. *Journal of biomechanics,* 32**,** 1349-1353.

Arampatzis, A., De Monte, G., Karamanidis, K., Morey-Klapsing, G., Stafilidis, S. & Brüggemann, G.-P. 2006. Influence of the muscle-tendon unit's mechanical and morphological properties on running economy. *Journal of Experimental Biology,* 209**,** 3345-3357.

Arampatzis, A., Morey-Klapsing, G., Karamanidis, K., Demonte, G., Stafilidis, S. & Brüggemann, G.-P. 2005. Differences between measured and resultant joint moments during isometric contractions at the ankle joint. *Journal of biomechanics,* 38**,** 885-892.

Askling, C., Karlsson, J. & Thorstensson, A. 2003. Hamstring injury occurrence in elite soccer players after preseason strength training with eccentric overload. *Scandinavian journal of medicine & science in sports,* 13**,** 244-250.

Astorino, T. A., Robergs, R. A., Ghiasvand, F., Marks, D. & Burns, S. 2000. Incidence of the oxygen plateau at VO2max during exercise testing to volitional fatigue. *Methods,* 3.

Avogadro, P., Kyröläinen, H. & Belli, A. 2004. Influence of mechanical and metabolic strain on the oxygen consumption slow component during forward pulled running. *European journal of applied physiology,* 93**,** 203-209.

Bergh, U., Sjodin, B., Forsberg, A. & Svedenhag, J. 1991. The relationship between body mass and oxygen uptake during running in humans.

Blickhan, R. 1989. The spring-mass model for running and hopping. *Journal of biomechanics,* 22**,** 1217-1227.

Boiteau, M., Malouin, F. & Richards, C. L. 1995. Use of a hand-held dynamometer and a Kin-Com® dynamometer for evaluating spastic hypertonia in children: a

reliability study. *Physical therapy,* 75**,** 796-802.
Bressel, E., Larsen, B. T., Mcnair, P. J. & Cronin, J. 2004. Ankle joint proprioception and passive mechanical properties of the calf muscles after an Achilles tendon rupture: a comparison with matched controls. *Clinical Biomechanics,* 19**,** 284-291.
Bressel, E. & Mcnair, P. J. 2001. Biomechanical Behavior of the Plantar Flexor Muscle-Tendon Unit after an Achilles Tendon Rupture No author or related institution has received any financial benefit from research in this study. *The American journal of sports medicine,* 29**,** 321-326.
Brughelli, M. & Cronin, J. 2008. A review of research on the mechanical stiffness in running and jumping: methodology and implications. *Scandinavian journal of medicine & science in sports,* 18**,** 417-426.
Buell, T., Green, D. & Risser, J. 1988. Measurement of the first metatarsophalangeal joint range of motion. *Journal of the American Podiatric Medical Association,* 78**,** 439-448.
Bunc, V., Heller, J. & Leso, J. 1988. Kinetics of heart rate responses to exercise. *Journal of sports sciences,* 6**,** 39-48.
Burr, D. B., Martin, R. B., Schaffler, M. B. & Radin, E. L. 1985. Bone remodeling in response to in vivo fatigue microdamage. *Journal of biomechanics,* 18**,** 189-200.
Butler, R. J., Crowell, H. P. & Davis, I. M. 2003. Lower extremity stiffness: implications for performance and injury. *Clinical Biomechanics,* 18**,** 511-517.
Caselli, M. A. & George, D. H. 2003. Foot deformities: biomechanical and pathomechanical changes associated with aging, Part I. *Clinics in podiatric medicine and surgery,* 20**,** 487-509.
Cavagna, G., Franzetti, P., Heglund, N. & Willems, P. 1988. The determinants of the step frequency in running, trotting and hopping in man and other vertebrates. *The Journal of physiology,* 399**,** 81-92.
Cavagna, G., Heglund, N. & Willems, P. 2005. Effect of an increase in gravity on the power output and the rebound of the body in human running. *Journal of Experimental Biology,* 208**,** 2333-2346.
Cavagna, G., Saibene, F. & Margaria, R. 1964. Mechanical work in running. *Journal of applied physiology,* 19**,** 249-256.
Cavagna, G., Willems, P., Franzetti, P. & Detrembleur, C. 1991. The two power limits conditioning step frequency in human running. *The Journal of physiology,* 437**,** 95-108.
Cavagna, G. A. 1975. Force platforms as ergometers. *Journal of applied physiology,* 39**,** 174-179.
Chelly, S. M. & Denis, C. 2001. Leg power and hopping stiffness: relationship with sprint running performance. *Medicine and Science in sports and Exercise,* 33**,** 326-333.
Cho, H.-Y., Sung In, T., Hun Cho, K. & Ho Song, C. 2013. A single trial of transcutaneous electrical nerve stimulation (TENS) improves spasticity and balance in patients with chronic stroke. *The Tohoku journal of experimental medicine,* 229**,** 187-193.
Cornu, C., Silveira, M.-I. A. & Goubel, F. 1997. Influence of plyometric training on the mechanical impedance of the human ankle joint. *European journal of applied physiology and occupational physiology,* 76**,** 282-288.
Dalleau, G., Belli, A., Bourdin, M. & Lacour, J.-R. 1998. The spring-mass model and

the energy cost of treadmill running. *European journal of applied physiology and occupational physiology,* 77, 257-263.

Dalleau, G., Belli, A., Viale, F., Lacour, J. & Bourdin, M. 2004. A simple method for field measurements of leg stiffness in hopping. *International journal of sports medicine,* 25, 170-176.

De Win, M. M., Theuvenet, W. J., Roche, P. W., De Bie, R. A. & Van Mameren, H. 2002. The paper grip test for screening on intrinsic muscle paralysis in the foot of leprosy patients. *International journal of leprosy and other mycobacterial diseases*, 70, 16-24.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I. & Steadman, J. R. 2003. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics,* 18, 662-669.

Delecluse, C., Van Coppenolle, H., Willems, E., Van Leemputte, M., Diels, R. & Goris, M. 1995. Influence of high-resistance and high-velocity training on sprint performance. *Medicine and Science in sports and Exercise,* 27, 1203-1209.

Doty, J. F. & Coughlin, M. J. 2013. Hallux valgus and hypermobility of the first ray: facts and fiction. *International orthopaedics,* 37, 1655-1660.

Dutto, D. J. & Smith, G. A. 2002. Changes in spring-mass characteristics during treadmill running to exhaustion. *Medicine and Science in sports and Exercise,* 34, 1324-1331.

Edouard, P., Codine, P., Samozino, P., Bernard, P.-L., Hérisson, C. & Gremeaux, V. 2013. Reliability of shoulder rotators isokinetic strength imbalance measured using the Biodex dynamometer. *Journal of Science and Medicine in Sport,* 16, 162-165.

Farley, C. T., Blickhan, R., Saito, J. & Taylor, C. R. 1991. Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits. *Journal of applied physiology,* 71, 2127-2132.

Farley, C. T., Glasheen, J. & Mcmahon, T. A. 1993. Running springs: speed and animal size. *Journal of Experimental Biology,* 185, 71-86.

Farley, C. T. & Gonzalez, O. 1996. Leg stiffness and stride frequency in human running. *Journal of biomechanics,* 29, 181-186.

Farley, C. T. & Morgenroth, D. C. 1999. Leg stiffness primarily depends on ankle stiffness during human hopping. *Journal of biomechanics,* 32, 267-273.

Ferber, R., Davis, I. M. & Williams Iii, D. S. 2003. Gender differences in lower extremity mechanics during running. *Clinical Biomechanics,* 18, 350-357.

Ferris, D. P., Louie, M. & Farley, C. T. 1998. Running in the real world: adjusting leg stiffness for different surfaces. *Proceedings of the Royal Society of London. Series B: Biological Sciences,* 265, 989-994.

Finley, J. M., Dhaher, Y. Y. & Perreault, E. J. 2012. Contributions of feed-forward and feedback strategies at the human ankle during control of unstable loads. *Experimental brain research,* 217, 53-66.

Günther, M. & Blickhan, R. 2002. Joint stiffness of the ankle and the knee in running. *Journal of biomechanics,* 35, 1459-1474.

Gajdosik, R. L., Allred, J. D., Gabbert, H. L. & Sonsteng, B. A. 2007. A stretching program increases the dynamic passive length and passive resistive properties of the calf muscle-tendon unit of unconditioned younger women. *European journal of applied physiology,* 99, 449-454.

Garth, W. P. & Miller, S. T. 1989. Evaluation of claw toe deformity, weakness of the foot intrinsics, and posteromedial shin pain. *The American journal of sports medicine,* 17**,** 821-827.

Goldmann, J.-P., Sanno, M., Willwacher, S., Heinrich, K. & Brüggemann, G.-P. 2013. The potential of toe flexor muscles to enhance performance. *Journal of sports sciences,* 31**,** 424-433.

Granata, K., Padua, D. & Wilson, S. 2002. Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks. *Journal of Electromyography and Kinesiology,* 12**,** 127-135.

Grimmer, S., Ernst, M., Günther, M. & Blickhan, R. 2008. Running on uneven ground: leg adjustment to vertical steps and self-stability. *Journal of Experimental Biology,* 211**,** 2989-3000.

Grimston, S. K., Engsberg, J. R., Kloiber, R. & Hanley, D. A. 2010. Bone mass, external loads, and stress fracture in female runners. *JAB,* 7.

Ha, S. C.-W., Fong, D. T.-P. & Chan, K.-M. 2015. Review of ankle inversion sprain simulators in the biomechanics laboratory. *Asia-Pacific Journal of Sports Medicine, Arthroscopy, Rehabilitation and Technology,* 2**,** 114-121.

Hashimoto, T. & Sakuraba, K. 2014. Strength Training for the Intrinsic Flexor Muscles of the Foot: Effects on Muscle Strength, the Foot Arch, and Dynamic Parameters Before and After the Training. *Journal of physical therapy science,* 26**,** 373-376.

He, J., Kram, R. & Mcmahon, T. A. 1991. Mechanics of running under simulated low gravity. *Journal of applied physiology,* 71**,** 863-870.

Heise, G. D. & Martin, P. E. 2001. Are variations in running economy in humans associated with ground reaction force characteristics? *European journal of applied physiology,* 84**,** 438-442.

Hicks, J. 1954. The mechanics of the foot: II. The plantar aponeurosis and the arch. *Journal of anatomy,* 88**,** 25.

Hobara, H., Muraoka, T., Omuro, K., Gomi, K., Sakamoto, M., Inoue, K. & Kanosue, K. 2009. Knee stiffness is a major determinant of leg stiffness during maximal hopping. *Journal of biomechanics,* 42**,** 1768-1771.

Hof, A., Geelen, B. & Van Den Berg, J. 1983. Calf muscle moment, work and efficiency in level walking; role of series elasticity. *Journal of biomechanics,* 16**,** 523-537.

Hopper, D. M., Grisbrook, T. L., Finucane, M. & Nosaka, K. 2014. Effect of ankle taping on angle and force matching and strength of the plantar flexors. *Physical Therapy in Sport,* 15**,** 254-260.

Hughes, J., Clark, P. & Klenerman, L. 1990. The importance of the toes in walking. *Journal of Bone & Joint Surgery, British Volume,* 72**,** 245-251.

Inman, V. T. & Eberhart, H. D. 1953. The major determinants in normal and pathological gait. *The Journal of Bone & Joint Surgery,* 35**,** 543-558.

Ishikawa, M., Komi, P. V., Grey, M. J., Lepola, V. & Bruggemann, G.-P. 2005a. Muscle-tendon interaction and elastic energy usage in human walking. *Journal of applied physiology,* 99**,** 603-608.

Ishikawa, M., Niemelä, E. & Komi, P. 2005b. Interaction between fascicle and tendinous tissues in short-contact stretch-shortening cycle exercise with varying eccentric intensities. *Journal of applied physiology,* 99**,** 217-223.

J. Stefanyshyn, D. & Nigg, B. M. 1998. Contribution of the lower extremity joints to mechanical energy in running vertical jumps and running long jumps. *Journal* *of sports sciences,* 16**,** 177-186.

Jones, A. M. 2002. Running economy is negatively related to sit-and-reach test performance in international-standard distance runners. *International journal of sports medicine,* 23**,** 40-43.

Jung, I.-G., Yu, I.-Y., Kim, S.-Y., Lee, D.-K. & Oh, J.-S. 2015. Reliability of ankle dorsiflexion passive range of motion measurements obtained using a hand-held goniometer and Biodex dynamometer in stroke patients. *Journal of physical therapy science,* 27**,** 1899.

Kemertzis, M. A., Lythgo, N. D., Morgan, D. L. & Galea, M. P. 2008. Ankle flexors produce peak torque at longer muscle lengths after whole-body vibration. *Medicine and Science in sports and Exercise,* 40**,** 1977-1983.

Kerdok, A. E., Biewener, A. A., Mcmahon, T. A., Weyand, P. G. & Herr, H. M. 2002. Energetics and mechanics of human running on surfaces of different stiffnesses. *Journal of applied physiology,* 92**,** 469-478.

Kim, D. Y., Park, C.-I., Chon, J. S., Ohn, S. H., Park, T. H. & Bang, I. K. 2005. Biomechanical assessment with electromyography of post-stroke ankle plantar flexor spasticity. *Yonsei medical journal,* 46**,** 546-554.

Kobayashi, T., Leung, A. K. & Hutchins, S. W. 2011. Design of a manual device to measure ankle joint stiffness and range of motion. *Prosthetics and orthotics international,* 35**,** 478-481.

Konrad, A., Gad, M. & Tilp, M. 2015. Effect of PNF stretching training on the properties of human muscle and tendon structures. *Scandinavian journal of medicine & science in sports,* 25**,** 346-355.

Krauss, I., Grau, S., Mauch, M., Maiwald, C. & Horstmann, T. 2008. Sex-related differences in foot shape. *Ergonomics,* 51**,** 1693-1709.

Kuitunen, S., Komi, P. V. & Kyröläinen, H. 2002. Knee and ankle joint stiffness in sprint running. *Medicine and Science in sports and Exercise,* 34**,** 166-173.

Kwah, L. K., Herbert, R. D., Harvey, L. A., Diong, J., Clarke, J. L., Martin, J. H., Clarke, E. C., Hoang, P. D., Bilston, L. E. & Gandevia, S. C. 2012. Passive mechanical properties of gastrocnemius muscles of people with ankle contracture after stroke. *Archives of physical medicine and rehabilitation,* 93**,** 1185-1190.

Lafortune, M. A., Henning, E. & Valiant, G. A. 1995. Tibial shock measured with bone and skin mounted transducers. *Journal of biomechanics,* 28**,** 989-993.

Lamontagne, A., Malouin, F. & Richards, C. L. 1997a. Viscoelastic behavior of plantar flexor muscle-tendon unit at rest. *Journal of Orthopaedic & Sports Physical Therapy,* 26**,** 244-252.

Lamontagne, A., Malouin, F. & Richards, C. L. 2000. Contribution of passive stiffness to ankle plantarflexor moment during gait after stroke. *Archives of physical medicine and rehabilitation,* 81**,** 351-358.

Lamontagne, A., Malouin, F., Richards, C. L. & Dumas, F. 1997b. Impaired viscoelastic behaviour of spastic plantarflexors during passive stretch at different velocities. *Clinical Biomechanics,* 12**,** 508-515.

Lastayo, P. C., Woolf, J. M., Lewek, M. D., Snyder-Mackler, L., Reich, T. & Lindstedt, S. L. 2003. Eccentric muscle contractions: their contribution to injury, prevention, rehabilitation, and sport. *Journal of Orthopaedic & Sports Physical Therapy,* 33**,** 557-571.

Lephart, S. M., Ferris, C. M., Riemann, B. L., Myers, J. B. & Fu, F. H. 2002. Gender differences in strength and lower extremity kinematics during landing.

Clinical orthopaedics and related research, 401**,** 162-169.
Lichtwark, G. A. & Wilson, A. 2008. Optimal muscle fascicle length and tendon stiffness for maximising gastrocnemius efficiency during human walking and running. *Journal of theoretical biology,* 252**,** 662-673.
Lindstedt, S. L., Reich, T. E., Keim, P. & Lastayo, P. C. 2002. Do muscles function as adaptable locomotor springs? *Journal of Experimental Biology,* 205**,** 2211-2216.
Luhtanen, P. & Komi, P. V. 1980. Force-, power-, and elasticity-velocity relationships in walking, running, and jumping. *European journal of applied physiology and occupational physiology,* 44**,** 279-289.
Maïsetti, O., Hug, F., Bouillard, K. & Nordez, A. 2012. Characterization of passive elastic properties of the human medial gastrocnemius muscle belly using supersonic shear imaging. *Journal of biomechanics,* 45**,** 978-984.
Mahieu, N. N., Mcnair, P., Cools, A., D'haen, C., Vandermeulen, K. & Witvrouw, E. 2008. Effect of eccentric training on the plantar flexor muscle-tendon tissue properties. *Medicine and Science in sports and Exercise,* 40**,** 117-123.
Malmir, K., Olyaei, G., Talebian, S. & Jamshidi, A. 2014. Viscoelastic response of the lateral side of the ankle to cyclic inversion: A time course analysis. *Scandinavian journal of medicine & science in sports,* 24**,** e477-482.
Mann, R. A. & Hagy, J. L. 1979. The function of the toes in walking, jogging and running. *Clinical orthopaedics and related research,* 142**,** 24-29.
Marsden, J., Ramdharry, G., Stevenson, V. & Thompson, A. 2012. Muscle paresis and passive stiffness: key determinants in limiting function in Hereditary and Sporadic Spastic Paraparesis. *Gait & posture,* 35**,** 266-271.
Matthiasdottir, S., Hahn, M., Yaraskavitch, M. & Herzog, W. 2014. Muscle and fascicle excursion in children with cerebral palsy. *Clinical Biomechanics,* 29**,** 458-462.
Mcmahon, T. A. & Cheng, G. C. 1990. The mechanics of running: how does stiffness couple with speed? *Journal of biomechanics,* 23**,** 65-78.
Mcmahon, T. A., Valiant, G. & Frederick, E. C. 1987. Groucho running. *Journal of applied physiology,* 62**,** 2326-2337.
Mcnair, P., Nordez, A., Olds, M., Young, S. W. & Cornu, C. 2013. Biomechanical properties of the plantar flexor muscle–tendon complex 6 months post-rupture of the achilles tendon. *Journal of Orthopaedic Research,* 31**,** 1469-1474.
Mcnair, P. J., Dombroski, E. W., Hewson, D. J. & Stanley, S. N. 2001. Stretching at the ankle joint: viscoelastic responses to holds and continuous passive motion. *Medicine and Science in sports and Exercise,* 33**,** 354-358.
Mcnair, P. J., Hewson, D. J., Dombroski, E. & Stanley, S. N. 2002. Stiffness and passive peak force changes at the ankle joint: the effect of different joint angular velocities. *Clinical Biomechanics,* 17**,** 536-540.
Menz, H. B. 2005. Analysis of paired data in physical therapy research: time to stop double-dipping? *Journal of Orthopaedic & Sports Physical Therapy,* 35**,** 477-478.
Mickle, K. J., Munro, B. J., Lord, S. R., Menz, H. B. & Steele, J. R. 2009. ISB Clinical Biomechanics Award 2009: toe weakness and deformity increase the risk of falls in older people. *Clinical Biomechanics,* 24**,** 787-791.
Miller, A. E. J., Macdougall, J., Tarnopolsky, M. & Sale, D. 1993. Gender differences in strength and muscle fiber characteristics. *European journal of applied physiology and occupational physiology,* 66**,** 254-262.

Miyazaki, S. & Yamamoto, S. 1993. Moment acting at the metatarsophalangeal joints during normal barefoot level walking. *Gait & posture,* 1**,** 133-140.
Moon, S. W., Kim, J. H., Jung, M. J., Son, S., Lee, J. H., Shin, H., Lee, E. S., Yoon, C. H. & Oh, M.-K. 2013. The effect of extracorporeal shock wave therapy on lower limb spasticity in subacute stroke patients. *Annals of rehabilitation medicine,* 37**,** 461-470.
Morin, J.-B., Dalleau, G., Kyrolainen, H., Jeannin, T. & Belli, A. 2005. A simple method for measuring stiffness during running. *J Appl Biomech,* 21**,** 167-180.
Morin, J., Samozino, P., Zameziati, K. & Belli, A. 2007. Effects of altered stride frequency and contact time on leg-spring behavior in human running. *Journal of biomechanics,* 40**,** 3341-3348.
Morita, N., Yamauchi, J., Kurihara, T., Fukuoka, R., Otsuka, M., Okuda, T., Ishizawa, N., Nakajima, T., Nakamichi, R. & Matsuno, S. 2014. Toe flexor strength and foot arch height in children. *Med Sci Sports Exerc*.
Moseley, A. & Adams, R. 1991. Measurement of passive ankle dorsiflexion: procedure and reliability. *Australian Journal of Physiotherapy,* 37**,** 175-181.
Mueller, M. J., Salsich, G. B. & Strube, M. J. 1997. Functional limitations in patients with diabetes and transmetatarsal amputations. *Physical therapy,* 77**,** 937-943.
Myerson, M. & Shereff, M. 1989. The pathological anatomy of claw and hammer toes. *The Journal of Bone & Joint Surgery,* 71**,** 45-49.
Nachbauer, W. & Nigg, B. M. 1992. Effects of arch height of the foot on ground reaction forces in running. *Medicine and science in sports and exercise,* 24**,** 1264-1269.
Nakamura, M., Ikezoe, T., Takeno, Y. & Ichihashi, N. 2011. Acute and prolonged effect of static stretching on the passive stiffness of the human gastrocnemius muscle tendon unit in vivo. *Journal of Orthopaedic Research,* 29**,** 1759-1763.
Nakamura, M., Ikezoe, T., Takeno, Y. & Ichihashi, N. 2012. Effects of a 4-week static stretch training program on passive stiffness of human gastrocnemius muscle-tendon unit in vivo. *European journal of applied physiology,* 112**,** 2749-2755.
Nesterenko, S., Domire, Z. J., Morrey, B. F. & Sanchez-Sotelo, J. 2010. Elbow strength and endurance in patients with a ruptured distal biceps tendon. *Journal of Shoulder and Elbow Surgery,* 19**,** 184-189.
Nightingale, E. J., Moseley, A. M. & Herbert, R. D. 2007. Passive dorsiflexion flexibility after cast immobilization for ankle fracture. *Clinical orthopaedics and related research,* 456**,** 65-69.
Nordez, A., Mcnair, P., Casari, P. & Cornu, C. 2009. The effect of angular velocity and cycle on the dissipative properties of the knee during passive cyclic stretching: a matter of viscosity or solid friction. *Clinical Biomechanics,* 24**,** 77-81.
Ogon, M., Aleksiev, A. R., Pope, M. H., Wimmer, C. & Saltzman, C. L. 1999. Does arch height affect impact loading at the lower back level in running? *Foot & ankle international,* 20**,** 263-266.
Peixoto, J. G., Dias, J. M. D., Dias, R. C., Da Fonseca, S. T. & Teixeira-Salmela, L. F. 2011. Relationships between measures of muscular performance, proprioceptive acuity, and aging in elderly women with knee osteoarthritis. *Archives of gerontology and geriatrics,* 53**,** e253-e257.
Porter, M. M., Andersson, M., Hellstrom, U. & Miller, M. 2002. Passive resistive torque of the plantar flexors following eccentric loading as assessed by

isokinetic dynamometry. *Canadian journal of applied physiology,* 27**,** 612-616.
Pousson, M., Van Hoecke, J. & Goubel, F. 1990. Changes in elastic characteristics of human muscle induced by eccentric exercise. *Journal of biomechanics,* 23**,** 343-348.
Prabhu, R., Swaminathan, N. & Harvey, L. A. 2013. Passive movements for the treatment and prevention of contractures. *Cochrane Database of Systematic Reviews*.
Putti, A., Arnold, G., Cochrane, L. & Abboud, R. 2007. The Pedar® in-shoe system: Repeatability and normal pressure values. *Gait & posture,* 25**,** 401-405.
Radin, E. L., Ehrlich, M. G., Chernack, R., Abernethy, P., Paul, I. L. & Rose, R. M. 1978. Effect of repetitive impulsive loading on the knee joints of rabbits. *Clinical orthopaedics and related research,* 131**,** 288-293.
Rao, S., Song, J., Kraszewski, A., Backus, S., Ellis, S. J., Md, J. T. D. & Hillstrom, H. J. 2011. The effect of foot structure on 1st metatarsophalangeal joint flexibility and hallucal loading. *Gait & posture,* 34**,** 131-137.
Rao, U. B. & Joseph, B. 1992. The influence of footwear on the prevalence of flat foot. A survey of 2300 children. *Journal of Bone & Joint Surgery, British Volume,* 74**,** 525-527.
Rapoport, S., Mizrahi, J., Kimmel, E., Verbitsky, O. & Isakov, E. 2003. Constant and variable stiffness and damping of the leg joints in human hopping. *Journal of biomechanical engineering,* 125**,** 507-514.
Reich, T., Lindstedt, S., Lastayo, P. & Pierotti, D. 2000. Is the spring quality of muscle plastic? *American Journal of Physiology-Regulatory, Integrative and Comparative Physiology,* 278**,** R1661-R1666.
Rizzo, J. A., Baker, D. I., Mcavay, G. & Tinetti, M. E. 1996. The cost-effectiveness of a multifactorial targeted prevention program for falls among community elderly persons. *Medical care,* 34**,** 954-969.
Roberts, T. J., Marsh, R. L., Weyand, P. G. & Taylor, C. R. 1997. Muscular force in running turkeys: the economy of minimizing work. *Science,* 275**,** 1113-1115.
Ross, S. A. & Engsberg, J. R. 2002. Relation between spasticity and strength in individuals with spastic diplegic cerebral palsy. *Developmental Medicine & Child Neurology,* 44**,** 148-157.
Selles, R. W., Li, X., Lin, F., Chung, S. G., Roth, E. J. & Zhang, L.-Q. 2005. Feedback-controlled and programmed stretching of the ankle plantarflexors and dorsiflexors in stroke: effects of a 4-week intervention program. *Archives of physical medicine and rehabilitation,* 86**,** 2330-2336.
Senda, M., Takahara, Y., Yagata, Y., Yamamoto, K., Nagashima, H., Tukiyama, H. & Inoue, H. 1999. Measurement of the muscle power of the toes in female marathon runners using a toe dynamometer. *Acta Med Okayama,* 53**,** 189-191.
Seyfarth, A., Blickhan, R. & Van Leeuwen, J. 2000. Optimum take-off techniques and muscle design for long jump. *Journal of Experimental Biology,* 203**,** 741-750.
Seyfarth, A., Geyer, H., Günther, M. & Blickhan, R. 2002. A movement criterion for running. *Journal of biomechanics,* 35**,** 649-655.
Singer, B., Dunne, J. & Singer, K. 2008. The short term effect of cyclic passive stretching on plantarflexor resistive torque after acquired brain injury. *Clinical Biomechanics,* 23**,** 1178-1182.
Sobolewski, E. J., Ryan, E. D. & Thompson, B. J. 2013. Influence of maximum range of motion and stiffness on the viscoelastic stretch response. *Muscle & nerve,*

48, 571-577.

Sobolewski, E. J., Ryan, E. D., Thompson, B. J., Mchugh, M. P. & Conchola, E. C. 2014. The influence of age on the viscoelastic stretch response. *The Journal of Strength & Conditioning Research,* 28, 1106-1112.

Stefanyshyn, D. J. & Nigg, B. M. 1997. Mechanical energy contribution of the metatarsophalangeal joint to running and sprinting. *Journal of biomechanics,* 30, 1081-1085.

Stefanyshyn, D. J. & Nigg, B. M. 1998. Dynamic angular stiffness of the ankle joint during running and sprinting. *Journal of applied biomechanics*, 14, 292-299.

Stel, V. S., Smit, J. H., Pluijm, S. M. & Lips, P. 2004. Consequences of falling in older men and women and risk factors for health service use and functional decline. *Age and ageing,* 33, 58-65.

Tanaka, T., Noriyasu, S., Ino, S., Ifukube, T. & Nakata, M. 1996. Objective method to determine the contribution of the great toe to standing balance and preliminary observations of age-related effects. *Rehabilitation Engineering, IEEE Transactions on,* 4, 84-90.

Taylor, G. & Allum, R. 1988. Ankle motion after external fixation of tibial fractures. *Journal of the Royal Society of Medicine,* 81, 19-21.

Taylor, M. J. & Beneke, R. 2012. Spring mass characteristics of the fastest men on earth. *International journal of sports medicine,* 33, 667.

Tilp, M., Steib, S. & Herzog, W. 2012. Length changes of human tibialis anterior central aponeurosis during passive movements and isometric, concentric, and eccentric contractions. *European journal of applied physiology,* 112, 1485-1494.

Tomassoni, D., Traini, E. & Amenta, F. 2014. Gender and age related differences in foot morphology. *Maturitas,* 79, 421-427.

Turner, D., Helliwell, P. S., Burton, A. K. & Woodburn, J. 2007. The relationship between passive range of motion and range of motion during gait and plantar pressure measurements. *Diabetic Medicine,* 24, 1240-1246.

Van Schie, C. H., Vermigli, C., Carrington, A. L. & Boulton, A. 2004. Muscle weakness and foot deformities in diabetes relationship to neuropathy and foot ulceration in Caucasian diabetic men. *Diabetes Care,* 27, 1668-1673.

Vattanasilp, W., Ada, L. & Crosbie, J. 2000. Contribution of thixotropy, spasticity, and contracture to ankle stiffness after stroke. *Journal of Neurology, Neurosurgery & Psychiatry,* 69, 34-39.

Viale, F., Dalleau, G., Freychat, P., Lacour, J.-R. & Belli, A. 1998. Leg stiffness and foot orientations during running. *Foot & ankle international,* 19, 761-765.

Wank, V., Frick, U. & Schmidtbleicher, D. 1998. Kinematics and electromyography of lower limb muscles in overground and treadmill running. *International journal of sports medicine,* 19, 455-461.

Whitting, J., Steele, J., Mcghee, D. & Munro, B. 2013. Passive dorsiflexion stiffness is poorly correlated with passive dorsiflexion range of motion. *Journal of Science and Medicine in Sport,* 16, 157-161.

Williams, D., Mcclay Davis, I., Scholz, J., Hamill, J. & Buchanan, T. 2003. Lower extremity stiffness in runners with different foot types. *Gait and posture*.

Williams, D. S., Davis, I. M., Scholz, J. P., Hamill, J. & Buchanan, T. S. 2004. High-arched runners exhibit increased leg stiffness compared to low-arched runners. *Gait & posture,* 19, 263-269.

Zhang, L.-Q., Chung, S. G., Bai, Z., Xu, D., Van Rey, E. M., Rogers, M. W., Johnson,

M. E. & Roth, E. J. 2002. Intelligent stretching of ankle joints with contracture/spasticity. *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* 10**,** 149-157.

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