FEET AND FOOTWEAR: FRIENDS OR FOES?

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A thesis submitted to the University of Birmingham for the degree of DOCTOR OF PHILOSOPHY
September 2017

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September 2017

Abstract

A third of over 65s have at least one fall per year whilst a quarter of over 45s endure foot pain. Footwear is associated with both fall risk and foot pain hence its investigation is of great importance. This thesis explores the potential benefits of minimalist footwear for the older adult population.

Chapter 2 ascertained the kinematic and kinetic differences between walking barefoot versus in footwear whilst highlighting the limited research on minimalist footwear, older adults and muscle activity differences. Accordingly, Chapter 3 outlined that minimalist footwear is kinematically more similar to barefoot, irrespective of age, thus offering a viable alternative. Similarly, Chapter 4 showed walking in minimalist footwear and walking unshod exhibit similar lower leg muscle activation patterns whilst differences exist to conventional footwear. Chapter 6 demonstrated how increasing intrinsic foot strength improved functional and static balance whilst Chapter 7 showed promise for minimalist footwear improving foot strength, functional balance, balance confidence as well as reducing foot and joint pain in a sample of older adults. In conclusion, this thesis highlights the need for future work to continue to investigate minimalist footwear in both older adults and other age groups for benefits to stability, foot health and joint pain.

Acknowledgements

I would first like to acknowledge my supervisors, François and Michael, who gave me the opportunity to do this PhD which has helped me to develop as both an academic and a person. I could not have completed this PhD without the help and friendship of numerous people and I will try my best to cover as many of these people as possible.

Firstly Borut, you helped me settle into the research group and office right from the first day and by the time you left you had become a great friend along with teaching me so much and in effect becoming my third supervisor! I tried to repay you with proof reading your endless supply of papers you impressively churned out and entertainment through my cycling failures and gymnastic attempts (!) but a huge thank you goes to you! Similarly Danique, I can't repay you enough for your Matlab help along the way! It was great to have you both in the research group (along with Laura on the occasions she was around!) but you all set very high standards to follow including you Laura who have always been a fraction better than me right from our undergrad!

Next I have to give a mention to the original great bunch of postgrads here when I first started. Benny, Fletchy, Ryan, Alex, Soden, Doris and Mike thanks for keeping me sane with our numerous barbeques (Fletchy and Ryan especially, Kings of the BBQ regardless of the time of year!), pool sessions (Benny, we need to add up the money and time we've spent around pool tables at Riley's and the Soak the past 4 years!), squash games and general good times! Most of you have finished up and moved on but of course Alex you are still plodding on as you do but maybe we can graduate together if you're not late to that as well!

To all of the newer bunch, Stu, Steve, Jamo, Stocksy, Gav, Ste, Nathan, Brandon, Seabright, Sarah, Sardar, again thanks for the support and friendship along the way and particularly in the case of Seabright and Stu endless hours of priceless entertainment! A special shout out also has to go to all the SportexR football team mates over the years, I won't name you all but you know who you are, definitely the best part of Mondays and Wednesdays if not the whole week. Save a space for me in next season's champions league as an honorary team member? I don't want to say goodbye just yet...

Finally from the PhD setting I would like to thank the techies Steve, Dave, Andy, Manjit and all the admin staff for all your help fixing any problem I encountered. Janice and Ned for providing additional support towards the end of my PhD; Anna, thanks for your Matlab help and I can't forget all of my volunteers who participated in my studies. I could not have done any of this without you and meeting and getting to know all of you during the data collection phases was the part of my PhD which I always enjoyed. I also need to thank the MRC-ARUK Centre, including Janet, Lisa and Alison, for the funding and support which enabled me to complete this PhD, Hobos Shoes and Two Barefeet Boarding Co. for the shoes I used as control shoes and Vivobarefoot for kindly providing the intervention shoes for my final study.

I also need to mention all my friends outside of my PhD life who have been there from the start and continue to offer their support and show interest in my work. Matt, you'll be glad to hear I'm finally growing up and leaving Uni! As much as you take the mick I know that you'll always be there if I need your help and you're truly a great friend. Gracie, your hard work and drive is inspiring and you deserve the very best from life. Ash and Tyler you're both just great guys with hearts of gold and I have no doubt we'll be friends for life.

Last and by no means least I need to thank my family and April for all the support you have given me throughout the last 4 years. It hasn't always been plain sailing but you were always there for me to encourage me to keep at it as you said it will all be worth it when I get to the end and look back. I'm glad I listened, I'm proud of the thesis I have produced as a result and I hope I've made you proud also. April thanks for putting up with me on a daily basis through it all, your approach to life is an inspiration to me, you always have a smile on your face and always have faith in me and what I can achieve. I couldn't have done it without you and without doubt the best part of my PhD was meeting you.

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%BW	percentage bodyweight
1MPJ	first metatarsophalangeal joint
3D	three dimensional
AAH	aquatic ape hypothesis
ABH	abductor hallucis
ADM	abductor digiti minimi
ANOVA	analysis of variance
AP	anterior-posterior
APA	anticipatory postural adjustments
BF	barefoot
BI	barefoot Indian
BLF	barefoot-like footwear (minimalist footwear)
BMI	body mass index
CNS	central nervous system
COM	centre of mass
CON	control shoes
COP	centre of pressure
CPG	central pattern generator
CS	control shoes
CSA	cross sectional area
DAQ	data acquisition device
EMG	electromyography
EVA	ethylene-vinyl acetate
FDB	flexor digitorum brevis
FDL	flexor digitorum longus
FHB	flexor hallucis brevis
FHL	flexor hallucis longus
FMM	foot mobility magnitude
FOF	fear of falling
FRT	functional reach test
g	grams
GCM	gastrocnemius
GRF	ground reaction force
gTFM	great toe flexor muscles
GTO	Golgi tendon organs
Hz	Hertz (frequency)
IDS	initial double support phase
kg	kilogram
LANK	left lateral malleolus
LDS	late double support phase

LHEE	left base of calcaneus
LKNE	left lateral epicondyle
LTOE	left 1st metatarsophalangeal joint
MG	medial gastrocnemius
MID	middle aged group
ML	medial-lateral
MSH	minimalist shoes (barefoot-like shoes)
mV	millivolts
MV	muscle volume
MVIC	maximum voluntary isometric contraction
N	Newtons
NHS	national health service
OLD	old aged group
OS	participants' own shoes
PCA	principal component analysis
PEA	prediction ellipse area
PL	peroneus longus
QP	quadratus plantae
RANK	right lateral malleolus
RASI	right anterior superior iliac spine
Res	resultant
RHEE	right base of calcaneus
RKNE	right lateral epicondyle
ROM	range of motion
RTOE	right 1st metatarsophalangeal joint
SD	standard deviation
SH	participants' own shoes
SI	shod Indians
SS	single support phase
TA	tibialis anterior
TFM	toe flexor muscles
TUG	timed up and go test
V	volts
VCR	vestibulocollic reflex
VOR	vestibulo-ocular reflex
VSR	vestibulospinal reflex
WS	western shod
YOUNG	young aged group

List of Publications

Chapter 2 – Franklin S, Grey M.J, Heneghan N, Bowen L, Li F.X. *Barefoot vs. common footwear: A systematic review of the kinematic, kinetic and muscle activity differences during walking,* Gait & Posture, 2015, **42**(3), 230-239

Chapter 3 – Franklin S, Grey M.J, Li F.X. *Is walking in barefoot-like shoes the same as walking barefoot. A kinematic and kinetic comparison across age.* Gait & Posture, 2017; **under review**

Chapter 4 – Franklin S, Li F.X, Grey M.J. *Modifications in lower leg muscle activation when walking barefoot or in minimalist shoes across different age-groups.* Gait & Posture, 2018, **60,** 1-5

Chapter 5 – Franklin S, Li F.X, Grey M.J. *Assessing the reliability of a customised device at measuring great toe flexor muscle (gTFM) strength and the importance of its study.;* **in preparation**

Chapter 6 – Franklin S, Li F.X, Grey M.J. *Increasing Foot Strength Improves Balance in Older Adults.;* **in preparation**

Chapter 7 – Franklin S, Grey M.J, Li F.X. *Does minimalist footwear improve balance and foot strength in older adults?;* **in preparation**

Chapter 1 – General Introduction

The American philosopher Eric Hoffer once said:

'The best part of the art of living is to know how to grow old gracefully.'

It is clear that growing old is not something which can be prevented, however living our older years out healthily and with grace is something we can strive for and is the focus of many researcher interests. A common consequence of ageing is an increased susceptibility to falls and injury as a result of age-related detriments in balance mechanisms. Maintaining balance is a multi-factorial process involving a combination of many sensorimotor mechanisms whose role is to provide continual feedback about the body's position and correct for these changes prior to reaching the outer limits of stability. There are many factors which can lead to an increased risk of instability including visual, vestibular and strength detriments as a result of ageing. This thesis aims to understand the contribution of the bottom up system; the proprioceptive and sensory feedback arising from the feet. The feet provide the base of support during walking and therefore can provide on-line feedback regarding the interaction of the body with the ground surface. As our body moves our feet have the inherent mechanisms present to interpret information such as pressure changes on the skin and the change foot shape due to the transfer of bodyweight providing information on our body position. This information is essential for movement control and balance modulation however can we better utilise these mechanisms? As modern humans we enclose our feet in footwear for protection from the environment in terms of hard or abrasive surfaces and to provide warmth. The question which this thesis aims to begin to answer is does wearing this footwear interfere with the sensitivity of the feedback mechanisms located within the foot, has it lead to reduced reliance on supportive musculature and does it impact upon the balance and gait ability of older adults.

1.1 Gait and Balance

1.1.1 The Gait Cycle

There is much debate throughout anthropologists about how and when humans became proponents of bipedal locomotion during evolution. One of the easiest theories to understand, but now widely disputed, is that of the aquatic ape hypothesis (AAH) which suggested that as our ancestors began living near a shoreline, wading in water was the trigger to begin to stand up in an upright posture and walk on two feet (Niemitz 2010). Whilst there were many suggested supporting factors put forward for the AAH by its authors these were primarily concerned with soft tissue and physiology developments. (Langdon 1997) explains how the increasing number of fossil discoveries do not support these suggestions with the fossil record demonstrating that key changes appear at different times and consequently cannot be the result of one significant change i.e. change in habitat.

Other more respected theories which have been put forward for being the origins of human bipedalism include the debate surrounding our evolutionary lineage originating from great apes, chimpanzees, bonobos and other non-human primates. These, whilst different, both have lifestyle feeding habits which may encourage pre-adaptation to bipedalism. Arboreal dwellers such as bonobos have been observed to exhibit sporadic displays of bipedalism whilst feeding (Stanford 2002). To reach fruit they stand upright on branches whilst using their arms as support and have been seen to walk bipedally from one feeding position to another without reverting to their normal quadrupedal position (Stanford 2002, Crompton 2016). On the other hand, terrestrial dwellers such as the non-human great apes although being primarily knuckle—walkers in a quadrupedal stance do adopt an upright posture when climbing trees exhibiting orthograde clambering behaviours akin to bipedalism (Crompton 2016). It is these aboreal origin theories which are supported by fossil evidence and may best explain our evolution to bipedalism

(Crompton 2016). However, the bipedal gait that we adopt today is still very different from our prebipedal ancestors and is a relatively recent evolutionary result (Schmitt 2003).

Human bipedal gait is described as an inverted pendulum where the centre of mass is propelled forwards in an arc with each step using a stiff-legged support (Alexander 1976). It comprises three main parts: an energy-dissipating and braking phase at initial contact using a transition from heel to sole; a vault over a stiff-legged flat footed support followed by a propulsive impulse through ankle plantarflexion at terminal stance (Usherwood, Channon et al. 2012). This type of gait is loosely described as a series of forward falls and recoveries and as such is especially economical due to the relatively small contribution of muscular force during the 'vault' phase (Usherwood, Channon et al. 2012). Upon step initiation the body effectively undergoes a forward fall by propelling the centre of mass in front of the base of support (Winter 1995). In order to prevent the fall, the opposite foot must be repositioned in front of the body and provide the base of support for the next step (Winter 1995). The initial contact of the repositioned foot occurs at the heel which means the ankle can then pivot from heel to sole effectively dissipating energy through, potentially, eccentric contraction of the tibialis anterior (TA) (Usherwood, Channon et al. 2012). With the foot flat and a relatively stiff leg, the body 'vaults' over its pivot with the ground reaction force (GRF) applied close to the ankle (Usherwood, Channon et al. 2012). This results in little lower leg muscle energy expenditure and hence high economy. The body then begins to undergo another forward fall and as the opposite foot is repositioned in front of the body the support foot becomes loaded at the forefoot in front of the ankle creating a moment arm. This enables the triceps surae to plantarflex the ankle increasing the effective leg length and providing a propulsive force (Usherwood, Channon et al. 2012). To terminate gait the centre of mass is kept within the base of support by applying braking forces through the support foot and no subsequent forward fall is initiated (Winter 1995, Whittle 2002).

This pattern of activity is under neural control via automatic behaviours originating from sensory receptors within the muscles and tendons of the legs. It has been shown in animals that a rhythmic sequence of alternating flexion and extension indicative of walking can be produced entirely within the spinal cord, termed the central pattern generator (CPG) (Duysens and Van de Crommert 1998). Whilst the rhythm can be generated by the spinal cord, the pattern is under constant modulation by higher brain centres and the sensory receptors (Duysens and Van de Crommert 1998). In humans it is still less clear whether this CPG is present however it is proposed that given the increased complexity of bipedal locomotion and the induced paralysis from spinal lesions it appears to indicate increased involvement of the corticospinal tract (Grey 2010). This will be explored further in a later section (1.1.2.2 Cutaneous and Proprioceptive) outlining the role of the proprioceptive system in walking and balance.

The recurring walking motion is termed the gait cycle and is commonly divided into two main phases: stance and swing (Figure 1). Stance is defined as the period when the foot is in contact with the ground and swing as the period when the foot is in the air. A normal walking gait will involve the initial contact with the ground occurring with the heel, termed heel strike, and at this point the primary goal is to reduce forward momentum. Immediately after heel strike is the bodyweight loading response where the foot has undergone plantar flexion and reached a position where it is flat. During this loading response the quadriceps, gluteals, tibialis anterior and peroneals serve to absorb the shock at the ankle and knee joints and transfer the load and stabilise the hip. Up until this point the body has been operating under conditions of double support with both feet in contact with the ground. Following this the other foot will begin to be lifted off the ground to initiate its swing phase and the body undergoes its first period of single leg support.

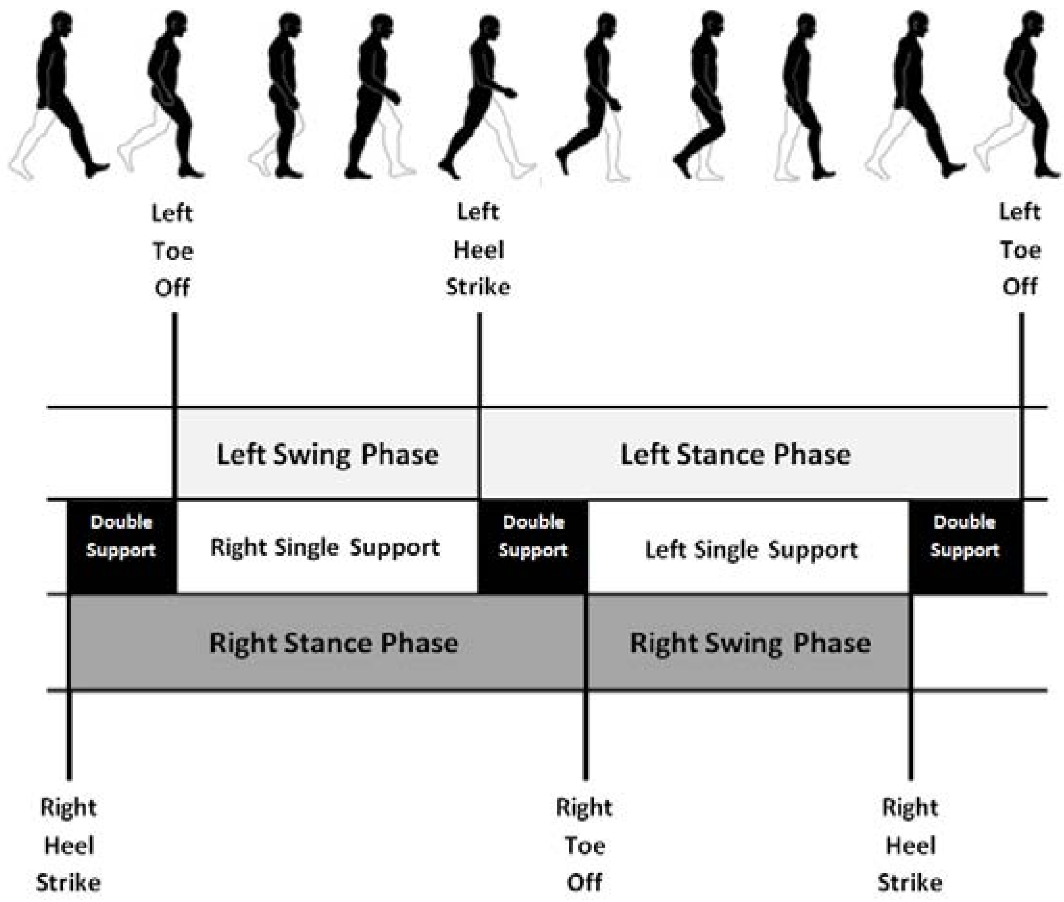

Figure 1: Displaying the phases of the gait cycle starting from right heel strike and finishing at second left toe off to encompass a full gait cycle for both the right and left legs. Adapted from (Perry and Burnfield 2010).

This point is termed mid-stance and is where the body undergoes its forward fall as the centre of mass travels anteriorly along the foot and eventually in front of the base of support. During mid-stance the triceps surae muscles are activated to control the forward motion of the tibia as the body pivots around the ankle joint. As the other leg is positioned in front of the body in order to 'catch' the fall and begin its stance phase, the body enters terminal stance and its second period of double support. At this point the body begins to be accelerated forward again and the heel begins to leave the ground through contraction of the ankle and toe plantar flexors. The final point of stance phase is termed "toe off" and is the point at which the toes leave the ground, begins its swing phase and the body enters its second period of single support. This action is produced by the ankle plantar flexors and the quadriceps.

The initial phase of swing involves rapid upward and forward acceleration as the foot is cleared from the ground through knee flexion via the quadriceps and ankle dorsiflexion by primarily the tibialis anterior. The mid-swing phase involves accelerating the leg so it is underneath the body before beginning deceleration as it begins to be extended out in front. The ankle dorsiflexors bring the foot back to a neutral position and the hamstrings act to begin deceleration of the leg. The final part is the terminal swing phase which involves rapid deceleration through hamstring activation and positioning of the leg and foot to prepare it for contact through quadriceps and ankle dorsiflexors activation (Kaufman and Sutherland 2005).

As displayed in Figure 2, in general the stance phase lasts for approximately 62% of the whole gait cycle and is subdivided into 5 sub divisions consisting of heel strike (0%), loading response (0-12%) , mid-stance (12-31%), terminal stance (31-50%) and toe off (50-62%). The swing phase on average lasts for approximately 38% and is subdivided into 3 sub divisions including initial swing (62-75%), mid-swing (75-87%) and terminal swing (87-100%) (Kaufman and Sutherland 2005). The durations of these phases however is purely dependant on the speed of walking with the relative swing phase duration increasing whilst the stance phase duration decreases with faster gait speed (Whittle 2002).

Other terms used during the analysis of gait include cadence, stride length, step length and step width. Cadence, or stride rate, is used to describe the number of steps in a given time and is usually referred to with units as steps per minute. Stride length refers to the distance covered in the direction of travel between two successive contacts of the same foot whilst step length uses the same principle but between successive contacts of the right and left feet. As an alternative definition, step length can has also been defined as the distance travelled by the trunk while a particular foot is on the ground (Alexander and Goldspink 1977). Step width most commonly defines the distance between consecutive contacts of the right and left feet but in the direction perpendicular to the direction of travel and as such is sometimes termed the base of support.

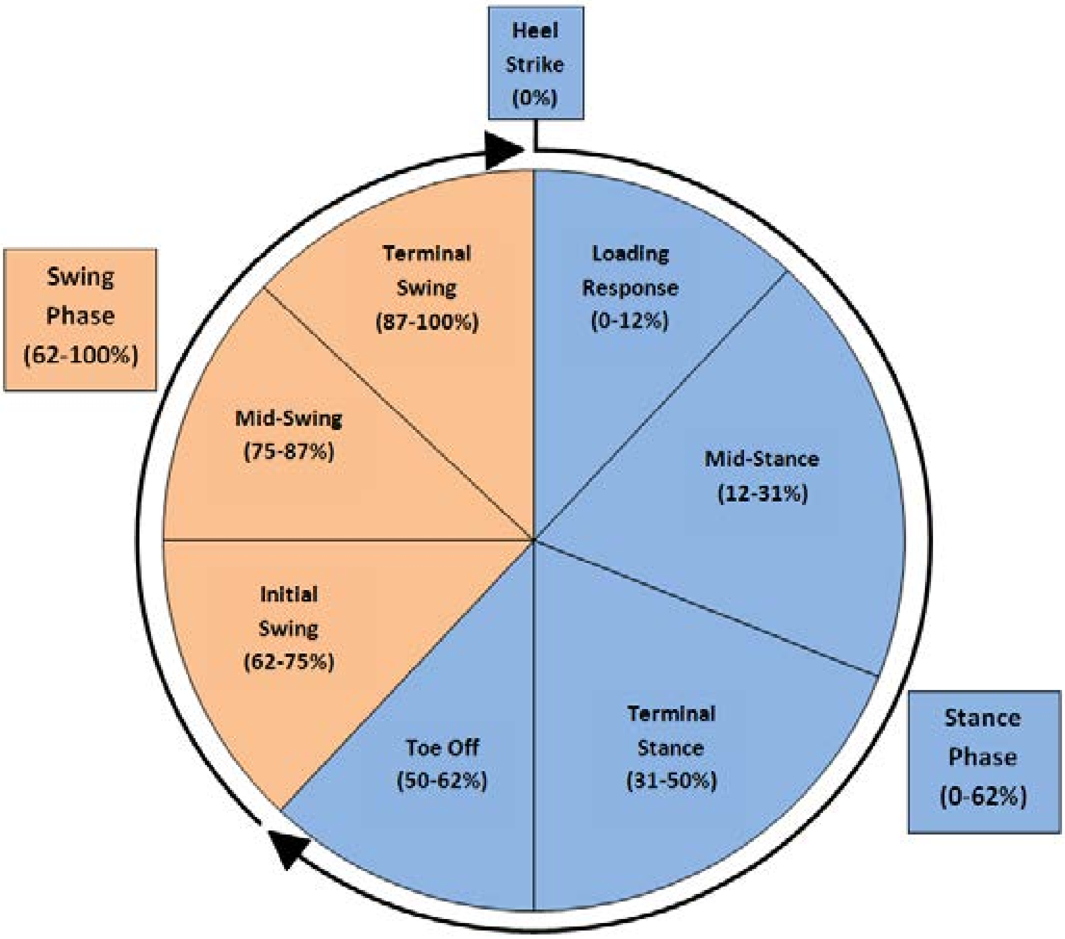

Figure 2 – Displaying the approximate timing intervals for the relative phases of one complete gait cycle using data from Kaufman et al. (Kaufman and Sutherland 2005) .

1.1.2 Balance Mechanisms

Balance, or stability, is defined as the ability to maintain the body's CoM above its base of support (BoS) and inside its limits of stability (Winter 1995). It involves the combination of a number of sensory and motor systems working together. Vision (eyes), vestibular (inner ear), touch (cutaneous afferents) and proprioception (muscle spindles and Golgi tendon organs) all provide information regarding how our body is positioned in relation to our environment. There is a degree of redundancy in the sensory system whereby if one sensory stream is lost or perturbed the remaining sensory systems can substitute the lost inputs and maintain functional balance (Winter 1995). In reality, due to the inverted pendulum characteristics of upright stance, we are never truly static and the system is under constant control by these mechanisms under conditions of dynamic stability. In these conditions the velocity of the CoM must also be accounted for along with its position with respect to the BoS and there is a degree in which the CoM can be outside of the BoS and balance can still be maintained (Hof, Gazendam et al. 2005). In effect, a combination of the position of the CoM and its velocity must be maintained within the margins of stability (the minimum distance from the CoM to the boundaries of the BoS) by the balance mechanisms in order to maintain dynamic stability (Hof, Gazendam et al. 2005).

The concept of dynamic stability is therefore concerned with the control of momentum of the CoM and becomes particularly relevant in more dynamic tasks such as walking (Meyer and Ayalon 2006). When walking the CoM is consistently outside of the BoS and both the CoM and BoS are in motion so to maintain stability, the relationship between the two trajectories must be carefully controlled. Instability results when there is a sudden disparity between the trajectories of the BoS and the CoM (Meyer and Ayalon 2006). The large relative mass of the head, arms and trunk means we are inherently unstable and controlling this during movement is of primary importance to prevent this disparity between BoS and CoM and associated imbalance. The central nervous system evokes postural changes to counteract any destabilising movement as a resultant of the large horizontal and lateral accelerations during walking minimising the movement of the head, arms and trunk (Winter 1995). Dynamic stability therefore involves not just maintaining stasis but also adjusting posture to cope with movement demands as well as reacting to perturbations (Meyer and Ayalon 2006).

1.1.2.1 Visual-Vestibular

Vision provides us with detail regarding the environment we are situated in and enables us to forward plan our movements to cope with demands placed upon us. The vestibular system, due to its location in the inner ear, provides us with information regarding linear and rotational motion of the head, and thus in turn the body, and is essential for maintaining body equilibrium. This is done via the two structures which make up the vestibular system; the otolith organs, which sense linear accelerations and our position relative to gravity, and the semicircular canals which sense rotation. The position and orientation of the semicircular canals and otolith organs in the ear enables detection of head movement in any direction. Both structures contain a gelatinous fluid, containing calcium carbonate crystals to increase its mass and inertia, which flows through the cavities and canals. Protruding into the fluid are hair cells which, as the fluid flows through the canals as a result of head movement, are bent in the direction the fluid is travelling. This bending of the hair cells causes them to depolarise or hyperpolarise depending on the direction evoking a neural response in the vestibular nerve. The vestibular output serves three important reflexes; the vestibulo-ocular reflex (VOR), the vestibulocollic reflex (VCR) and the vestibulospinal reflex (VSR)(Hain and Helminski 2007). The VOR enables vision to remain stable through periods of head motion by generating compensatory eye movements. Similarly, the VCR stabilises the head by acting on the neck muscles. The VSR is responsible for producing compensatory body movements in order to maintain postural stability and prevent falling. Each of these vestibular reflex pathways involve processing through the central nervous system (CNS) where the motor output is produced through higher cortical areas and adjusted if necessary based on previous experiences by the cerebellum (Hain and Helminski 2007).

1.1.2.2 Cutaneous and Proprioceptive

Touch/cutaneous and proprioceptive information provide us with information about how our body segments are positioned in relation to each other and also how our body is positioned in relation to the environment. The information arises from a range of areas including sensory cutaneous receptors which are sensitive to pressure and stretch, muscle spindles which are sensitive to changes in muscle length and velocity and Golgi tendon organs which are sensitive to muscle tension and force.

In terms of maintaining balance in upright stance the sensory cutaneous receptors on the soles of the feet provide information regarding where the body's centre of mass is situated with respect to the base of support (Roll, Kavounoudias et al. 2002). The cutaneous receptors are sensitive to pressure changes and therefore as the body sways in upright stance, given its characteristics similar to that of an inverted pendulum, the movement of the mass evokes pressure differentials along the surface of the foot sole (Kavounoudias, Roll et al. 1998). This pressure change stimulates the receptors and via an automatic behaviour, efferent pathways are triggered to counteract the sway and maintain stability (Kavounoudias, Roll et al. 1998). These foot sole inputs work in a complementary fashion with proprioceptive information from the ankle muscles (described below) to help maintain balance (Kavounoudias, Roll et al. 2001). Researchers have investigated the role of cutaneous afferents by either facilitating or inhibiting their action. Recent studies have showed that plantar desensitisation through cold water immersion of the feet impairs balance and/or gait modulation (Nurse and Nigg 2001, Lin and Yang 2011). Some studies have reported that increasing the plantar cutaneous information via vibration or textured insoles can serve to reduce sway and gait variability implying greater balance control (Maki, Perry et al. 1999, Kavounoudias, Roll et al. 2001, Priplata, Niemi et al. 2003, Galica, Kang et al. 2009) however there is still some debate around their benefits with other studies demonstrating no benefit to healthy older people (Hatton, Dixon et al. 2009, Hatton, Dixon et al. 2012) or patients with neurological disorders (Kalron, Pasitselsky et al. 2015). It has been shown however that older adults appear to rely more on this cutaneous information to maintain balance than younger adults (Machado, da Silva et al. 2017).

Mechanoreceptors located in the muscles provide direct information regarding the positions of our limbs and the forces generated by them. There are two main types of mechanoreceptors namely the muscle spindles and the Golgi tendon organs (GTO) (Carpenter 2003). The muscle spindles are situated within each muscle and their fibres, denoted intrafusal fibres, run parallel to the contractile muscle fibres or extrafusal fibres (Carpenter 2003). There are more muscle spindles located in regions where precise movement control is required such as the neck, hands and feet (Grey 2010). Wrapped around the intrafusal fibers are one primary and a number of secondary endings of Group Ia and Group II afferent sensory neurons respectively (Grey 2010). The intrafusal fibre location means they are perfectly situated to record changes in muscle length as they are stretched simultaneously exciting both sensory endings. Whilst secondary endings simply increase their firing monotonically with the degree of muscle stretch, primary fibres respond dynamically indicating the velocity and acceleration of the stretch (Grey 2010). This change in firing rate is relayed into the CNS and results in a muscle contraction to control the rate of change in length (Carpenter 2003). Their role in balance control is highlighted due to their direct interaction with the CNS and in particular the cerebellum(Marks 2015). Of great importance is the ability of the muscle spindles to modulate their sensitivity as muscle length changes. Clearly as muscle length changes, via alpha motor neuron activation, the length of the muscle spindles needs to change accordingly to ensure that the sensitivity of these spindles is recalibrated to this new length. This is done via the efferent portion of the fusimotor system (with the muscle spindles being the afferent portion), namely gamma motor neurons. It was thought that whilst alpha motor neurons control the length of muscle fibres, gamma motor neurons simultaneously adjust the length of the muscle spindles hereby ensuring sensitivity to stretch across all muscle lengths. This process was defined as alpha gamma co-activation (Carpenter 2003).

This afferent feedback mechanism is essential for upright stance and provides the basis for the control of human standing. As previously stated the body acts like an inverted pendulum, and when under quiet stance the ankle joint serves as the pivot. The main method used to maintain balance during normal quiet standing is the ankle strategy whilst a second method employed during large perturbations or with a narrow base of support is the hip strategy. The ankle strategy involves actively applying small corrections to counteract the direction and magnitude of sway as sensed by length changes in the muscles crossing the ankle joint. There is much debate as to whether this operates as a feedback loop through the mechanoreceptors sensing a change and activating a muscular response in response to it or if in fact there is feed-forward control where the CNS is effectively predicting future body position based on prior position change knowledge and activates a muscular response in preparation (Masani, Popovic et al. 2003). Nonetheless, the muscle spindle afferent information contributes to sensing changes in body position and eliciting a postural correction.

The muscle proprioceptors are also involved in ensuring balance whilst walking. The primary locomotor pattern is generated within a spinally mediated pathway called central pattern generators and afferent information is essential in modulating this output. One role the muscle proprioceptors play during walking is to monitor the loading of the legs to ensure effective phase transition from stance to swing such that adequate support is provided to the body prior to initiating the swing phase (Duysens, Clarac et al. 2000, Lam and Pearson 2002, Pearson 2004). It is thought that the loading is monitored by the muscle spindles and GTO (Duysens, Clarac et al. 2000). The GTO are receptive to changes in tension or load and hence during active movement through concentric muscle contraction they provide similar information to that of the muscle spindles due to an increase in tension and a decrease in muscle length (Carpenter 2003). They can also be receptive to passive movements due to a change in load and tension. As with the muscle spindles the GTO modulate muscle activity via automatic behaviours from the afferent motor neurons, through inter-neurons within the spinal cord and via efferent motor neurons to the muscles (Lam and Pearson 2002). This information serves as an input to the central pattern generator to modulate its output. It is generally thought however that afferent information is also involved supraspinally in reorganizing cortical networks and adaptive responses to perturbations (Pearson 2004). When load is applied to the muscle the GTO afferent firing frequency is a signal of the force being produced. Predominantly the GTO afferent firing has an inhibitory action on muscle activity however it has been shown during locomotion to reverse and have an excitatory input on the ankle extensors (Pearson 1995). It is suggested that the GTO force feedback directly modulates ankle extensor activity to ensure that stance is maintained whilst ankle extensors are undergoing loading (Pearson 1995).

In terms of muscle spindle involvement during locomotion, it is clear their location within each locomotor muscle, their superior connections to the central nervous system and their ability to modulate their sensitivity to muscle length make them ideally suited to controlling and/or adapting locomotion. The premise of alpha gamma co-activation was introduced earlier however in terms of muscle spindle involvement in locomotion this principle has begun to be disputed. As more research was designated to this area it was discovered that there were different types of gamma motor-neurons, dynamic and static, which innervate different types of spindle afferents, secondary and primary. It’s suggested that static gamma motor-neurons serve to innervate muscle spindle secondary endings whilst dynamic gamma motor-neurons innervate muscle spindle primary endings. Interestingly the gamma motor-neurons have been shown to display distinctly different patterns of activation to alpha motor neurons therefore suggesting that alpha gamma co-activation may not solely explain the role of the fusimotor system (Ellaway, Taylor et al. 2015). The fact that the muscle spindle system is the only sensory feedback structure to be controlled directly by the CNS highlights its respective importance and superior role (Ellaway, Taylor et al. 2015). It was proposed that during locomotion the fusimotor drive, primarily through the static gamma motor-neurons, may innervate the muscle spindle secondary endings to inform the central nervous system of the intended movement and effectively provide a 'temporal template' for the locomotor action. Concurrently, the dynamic gamma motor-neurons prime the muscle spindle primary afferents to detect any deviations from the intended plan and to signal phase transitions via changes in muscle lengthening (Ellaway, Taylor et al. 2015). In this arrangement the muscle spindles are effectively serving to plan and correct the locomotor pattern simultaneously under direct control from the CNS. It has to be stated however that due to the difficulty of studying muscle spindles directly this research is purely from fully intact or reduced (decerebrate) cat models as opposed to human. Human studies have primarily been limited to perturbing the proprioceptors through means of vibration to elucidate their role in locomotion.

1.1.3 The Role of the Foot

The foot is comprised of 26 bones, 33 joints and 19 muscles which are structured in a way in which it can adapt to the demands placed upon it. The foots primary role is to provide the base of support for our upright bipedal stance; this requires an ability to remain stiff under periods of force transmission whilst also being flexible at periods of load absorption and having the ability to conform to differing surface characteristics. The principal component of the foot structure which enables these qualities is the arrangement of the bones into an arch running along the longitudinal length of the foot. This structural formation allows the foot to compress and flatten during periods of load absorption whilst remaining strong and structurally sound during force transmission. The plantar aponeurosis, which is the elastic tendinous tissue which spans the sole of the foot running from the base of the calcaneus to the heads of the metatarsals, assists with the maintenance of this structure through the storage of elastic energy under stretch when the arch compresses. This energy can then be released when the load is removed to assist in returning the foot's structure to its prior state and in transmitting forces to the forefoot to provide efficient propulsion(Erdemir, Hamel et al. 2004).

From an anatomical and biomechanical perspective the foot plays a particularly important role in the final phase of gait and in fact makes up one of the six determinant of gait initially proposed by Saunders et al. in the 1950's (Whittle 2002). As the body enters the terminal stance phase the ankle moves from a period of dorsiflexion into plantarflexion as the heel rises from the ground (Whittle 2002). The forefoot and toes continue to remain in contact with the ground with the action of the ankle plantar flexors and as such serves to increase the effective leg length. This increase in effective leg length helps to stabilise the body by reducing the centre of mass vertical excursion (Whittle 2002).

There is an increasing amount of research which is indicating how the muscles in the foot contribute to gait and balance. The muscles responsible for foot motion are divided into intrinsic or extrinsic muscles. The extrinsic muscles are external to the foot in origin whereby they have an action on the foot and toes via tendons but the majority of the muscle itself is situated in the shank e.g. flexor digitorum longus (FDL) and flexor hallucis longus (FHL). The FHL has been shown to contribute during the propulsive phase of gait and particularly during faster walking speeds (Peter, Hegyi et al. 2015). The relative contribution during isometric plantarflexion is considerably less than during walking highlighting the task-dependant functionality of the FHL being primarily during locomotion (Peter, Hegyi et al. 2015). During the terminal stance phase the force on the first metatarsal head was estimated to be as much as 119% of body weight with 52% of bodyweight being exerted through the FHL tendon (Jacob 2001). Aside from providing force to the ground for propulsion the action of the FHL along with the other toe flexors and ankle stabilisers such as peroneus longus (PL) serve to support the longitudinal arch and maintain structural integrity during this phase of force transmission (Jacob 2001).

The smaller intrinsic foot muscles are situated entirely within the foot itself, examples being, flexor digitorum brevis (FDB) and abductor hallucis (ABH). These smaller muscles have also been shown to offer active support to the longitudinal arch to assist the plantar aponeurosis in maintaining foot structural integrity (Fiolkowski, Brunt et al. 2003), assist in load absorption (Kelly, Cresswell et al. 2014) and contribute to efficient force transmission to the ground (Kelly, Lichtwark et al. 2015). Quadratus plantae (QP), originating from the medial and lateral sides of the calcaneus and inserting into the FDL tendons, is unique to the human foot and is suggested to assist the FDL during bipedal locomotion (Soysa, Hiller et al. 2012). Due to its location the action of the FDL would be to move the toes medially; it is thought that the QP contracts concurrently to redirect the FDL contraction to produce flexion (Soysa, Hiller et al. 2012). These muscles therefore act in a synergistic fashion using the force from the larger FDL and the location of the QP to evoke the correct action necessary for upright walking.

The intrinsic foot muscles are perfectly situated for stability by providing support and helping maintain a rigid structure through which the larger force producing muscles can act through. They may also be important in increasing the contact area at push off by controlling dorsiflexion at the metatarsophalangeal joints ensuring toes remain flat against the ground (Soysa, Hiller et al. 2012). This would also have the action of improving pressure distribution across the metatarsals and reducing localised impulses (Soysa, Hiller et al. 2012).

The foot muscles have also been suggested to contribute to maintaining balance with weakness in these muscles highlighted as a precursor to increased fall risk (Mickle, Munro et al. 2009). Hallux plantar flexion strength is deemed an important determinant of functional balance ability in older adults (Spink, Fotoohabadi et al. 2011). Their active role in maintaining balance was demonstrated by an increase in activation of the plantar intrinsic muscles during periods of increased postural demand. There was a strong correlation witnessed between the EMG amplitude and medial-lateral centre of pressure deviations during single leg stance (Kelly, Kuitunen et al. 2012). It seems clear that weakness within these muscles would have implications on both gait performance through reduced structural integrity of the foot, namely the function of the longitudinal arch, resulting in impaired load absorption and force transmission as well as maintaining stability during functional balance tasks and periods of increased postural demand.

1.2 An Ageing Population

1.2.1 Population Statistics

Data from the Office of National Statistics indicates the UK population stood at 64.6 million in 2014 with 17.7% aged over 65 which equates to approximately 11.4million people. The projected population in 2039 will stand at 74.3 million with 24.3% of these aged over 65 equating to approximately 18 million people (ONS 2016). Whilst this projected increase in overall population includes rises in migration, the remaining increase in number is due to the births to deaths ratio with people living considerably longer. This is resulting in an ageing population with the life expectancy of a baby born in 2016 being 90.6 years for a boy and 93.5 years for a girl with a continuing trend to increase (ONS 2016). This clearly has an economic impact regarding the length of time pensions are required to last for but also of relevance is the increased reliance placed on the health service. The Department of Health highlighted this with the statistic that people over the age of 65 years old account for approximately 40% of all hospital bed days and 65% of the NHS spending budget is spent on this age group (Health 2010). The main problem lies in that the overall life expectancy is rising faster than the disability-free life expectancy (Mortimer and Green 2015). This means that even though we are living longer, this extra time in older age is spent with one or multiple detrimental health conditions such as frailty, coronary heart disease and dementia which require assistance and support from health services and/or associated charities. It is therefore of major importance to find ways of reducing this prevalence of ill health and maximise the number of years spent healthy and independent in older age.

According to data from both the US Health Retirement Survey and the English Longitudinal Study of Ageing the most common health conditions experienced in older age include heart disease(10-15% of the population sample), hypertension(34-42%) and diabetes(6-13%) (Banks, Marmot et al. 2006). Furthermore, diabetes prevalence is growing considerably and is thought to increase by more than double worldwide by 2030 with the largest rise in the over 65 years old age group(Christensen, Doblhammer et al. 2009). However, there is extensive research to show beneficial effects of physical activity at reducing the risk of these health conditions and in the management of these conditions (Anderson, Oldridge et al. 2016, Kyu, Bachman et al. 2016). A worldwide study on the burden of disease and life expectancy reported physical inactivity to be the fourth most important risk factor in the UK with it contributing to one in every six deaths (Lee, Shiroma et al. 2012). It has been reported that only 30% of people are achieving the recommended levels of physical activity in every decade of their life with a steady decline in physical activity being observed from the age of 50 onwards (Wright, Robinson et al. 2012). It is therefore clear that researching ways to increase physical activity levels across the whole population but particularly those in or approaching older age is of vital importance.

1.2.2 Age-Related Sensory and Motor Deficits

In the previous section the mechanisms through which balance is maintained was discussed; the visual, vestibular, touch and proprioceptive sensory mechanisms. As we age these mechanisms undergo a decline in their sensitivity and this can have a detrimental impact on balance as well as the ability to perform certain activities of daily life.

1.2.2.1 Visual

There are three main visual problems which impact on the functional ability in older adults; spatial contrast sensitivity deficit, scotopic function decline and slower visual processing speed(Owsley 2016). Spatial contrast sensitivity is defined as the ability to discriminate between light and dark and how small the difference between them can be perceived and is important in object identification and reading. The contrast sensitivity is reduced in ageing due to the reduction in pupil size affecting the optical properties as well as the increased development of a crystalline lens which leads to cataracts(Owsley 2016). Scotopic visual function is vision under low light conditions and a commonly reported problem in older adults is a reduced speed of dark adaption leading to visual problems at night. The mechanism thought to explain this issue is of metabolic origin being a decrease in the speed of rhodopsin regeneration (Owsley 2016). Visual processing speed decline is common in older age with approximately 25-30% displaying a slowing and it is attributed to inefficiencies in visual searching and attentional control (Owsley 2016). All three of these declines in visual ability have been linked with an increased risk of falls, motor vehicle collisions and problems performing everyday tasks due to a slowing in the detection of visual cues and the subsequent reaction.

1.2.2.2 Vestibular

The role of the vestibular system as previously mentioned is to maintain body equilibrium and to provide a sense of head orientation with respect to gravity. As a result of ageing the sensitivity of this mechanism can be affected and balance ability impaired. It is commonly reported that as we age we undergo vestibular hair cell loss, the cells which sense the acceleration and cause depolarisation of the vestibular nerve, as well as vestibular sensory neurons within the nuclei and cerebellum and vestibular nerve fibres themselves (Anson and Jeka 2015, Zalewski 2015). The loss of sensory cells leads to a reduced sensitivity to head acceleration and this has been suggested to lead to impairments in gaze stability and postural compensatory reactions during body motion via the vestibular evoked reflexes (Anson and Jeka 2015). Around the ages of 70-80 years old is when the loss of hair cells is shown to be most prominent and this corresponds with the time in which complaints about balance ability detriments, confidence and dizziness arise (Zalewski 2015).

1.2.2.3 Cutaneous and proprioceptive

There also appears to be age-related changes to touch and proprioceptive sensitivity. It has been shown that older adults have reduced tactile sensitivity with a significantly greater threshold to light touch, vibration and reduced spatial acuity than their younger counterparts (Perry 2006, Wickremaratchi and Llewelyn 2006). The reduction in tactile acuity has also been reported to be the most prominent in the distal regions such as the hands and feet compared to more proximal regions like the forearm(Wickremaratchi and Llewelyn 2006). The reason for this in unclear however it has been established that it is not purely a result of greater physical wear and tear in these contact regions(Stevens, Alvarez-Reeves et al. 2003). Functionally, this decrease in tactile sensitivity has been shown to be a strong predictor of postural instability in older age and in particular pressure feedback from the great toe region(Tanaka, Noriyasu et al. 1996). Significant correlations have been demonstrated between plantar cutaneous vibratory detection thresholds and postural sway amplitude and frequency (Peters, McKeown et al. 2016). There are also recent research studies suggesting that both upper (Herter, Scott et al. 2014) and lower (Petrella, Lattanzio et al. 1997, Relph and Herrington 2016) limb position sense appears to worsen with age. In these studies position sense accuracy was determined through position matching tasks of either the contralateral limb or via repeated movements of the ipsilateral limb and the error between them recorded. Older adults were considerably worse at position matching than the younger adults however regular activity appears to help to preserve position sense with considerably less matching error in those elderly participants who were habitually active (Relph and Herrington 2016). This has been supported by a physical training study on elderly women which showed that proprioception, as determined by position matching as well as movement detection threshold, was improved following 6 weeks of regular activity (Thompson, Mikesky et al. 2003). This study included 2 exercise groups, one which underwent resistance training and one which completed the same activities through the same range of motion but with no loading. Of note is that both groups experienced similar significant improvements in dynamic and static proprioception highlighting that the proprioceptive improvement is independent of muscle strength/muscle loading. It is thought that regular exercise and in particular repetitive motor movements, which repeatedly activate the muscle spindle afferents, leads to improved proprioceptive sensitivity via increased gamma motoneuronal activity and a subsequent increased reliance on afferent information (Thompson, Mikesky et al. 2003, Relph and Herrington 2016).

1.2.2.4 Muscle Strength

There is an abundance of literature demonstrating an age-related decline in muscle strength. The reasons for this decline in muscle strength were often put down to sarcopenia; the age-associated loss of skeletal muscle mass. What is also now known however is that sarcopenia per se does not fully account for the decline in strength but muscle quality is also affected in the ageing process (Goodpaster, Park et al. 2006, Delmonico, Harris et al. 2009). In the Health, Aging and Body Composition Study it was shown that there was an age-related increase in intramuscular fat regardless of whether the participants remained weight stable or not over the 5 year period (Delmonico, Harris et al. 2009). This compromise in muscle quality must also account for a proportion of the muscle strength detriments witnessed with ageing as the decreases in strength were 2-5 times greater than the loss in muscle size due to sarcopenia (Delmonico, Harris et al. 2009). The majority of these studies have understandingly focussed on the large lower leg muscles such as the quadriceps or in functional upper body muscles such as the biceps for their relevance in activities of daily independent living. It is of paramount importance to ensure that muscle strength remains at a level whereby older adults remain able to complete tasks such as raising from a chair, climbing stairs and holding a kettle for as long as possible such that they can remain independent. As such interventions targeted at maintaining or improving muscle strength in older adults are of interest. Encouragingly research in these areas has demonstrated that older adults do remain responsive to strength increases with training and the age-related declines in strength can be slowed and reversed. One example of this was that a 10 week moderate intensity resistance strength training program in older aged men (61-75years) demonstrated a 32% and 28% increase in knee extension and knee flexion strength respectively (Kalapotharakos, Smilios et al. 2007). This highlights how remaining physically active in later life can help to attenuate the detrimental effects of ageing.

Research has also explored smaller muscles such as those within the feet. Using ultrasound to visualise the quantity and quality of the ABH muscle in three age groups (20-44years, 45-64years and 65+years) with hallux valgus it was found that there was a significant reduction in muscle thickness and cross sectional area in the older age group compared to the 20-44year old age group suggesting muscle atrophy (Aiyer, Stewart et al. 2015). Interestingly there was no significant difference in muscle quality in this muscle between age groups as determined by echo-intensity measures (Aiyer, Stewart et al. 2015). This data is supported by another ultrasound study which assessed the size of multiple foot muscles namely abductor hallucis (ABH), flexor hallucis brevis (FHB), flexor digitorum brevis (FDB), quadratus plantae (QP) and abductor digit minimi (ADM) as well as flexor digitorum longus (FDL) and flexor hallucis longus (FHL) situated in the shank (Mickle, Angin et al. 2016). Alongside muscle quantity measures the authors also directly assessed toe plantar flexor strength between their young and old participants and witnessed a 38% and 35% reduction in hallux and lesser toe strength respectively within the older age group (Mickle, Angin et al. 2016). This coincided with a significant reduction (19%-45%) in thickness and cross sectional area of all muscles aside from the ABH and FHB further indicating how muscle strength and size is decreased with advanced age (Mickle, Angin et al. 2016). Likewise with the larger muscles it also appears that the smaller muscles within the feet can be trained and this reduction in strength and size can be diminished. Increases in toe flexor strength have been witnessed following a resistance training program in both young (~50% increase in toe grip force) (Hashimoto and Sakuraba 2014) and old (~ 36% increase in peak toe flexor force) (Mickle, Caputi et al. 2016) participants but although encouraging, this research is limited in comparison to the training of larger muscles like the quadriceps.

1.2.3 Foot Problems and Foot Pain

Older adults also have a high prevalence of foot problems and associated foot pain which can often restrict mobility and physical activity participation (Menz, Dufour et al. 2013). It has been reported that hallux valgus, or bunions, are present in 33-44% of women and 14-25% of men and lesser toe deformities including curled or hammer toes present in 37-76% of women and 54-59% of men depending on the study and population sample (Dawson, Thorogood et al. 2002, Dunn 2004, Menz and Morris 2005, Nix, Smith et al. 2010). Arthritis is also a common problem within the feet of older adults and a recent radiographic study on over 9000 adults >50 years old reported that osteoarthritis was present in 1 in 6 with the vast majority of these also reporting disabling foot symptoms (Roddy, Thomas et al. 2015). There is much research into what the causes are of these foot problems in older age but the general consensus appears to suggest the primary risk factors are ill-fitting footwear (Burns, Leese et al. 2002, Menz and Morris 2005, Dufour, Broe et al. 2009), foot muscle weakness (Whitman 2010) and being overweight (Nguyen, Hillstrom et al. 2010) or obese (Mickle and Steele 2015). Incorrectly fitting footwear is a common issue within older age with up to 78.4% wearing shoes which are narrower than their foot width and 10.2% wearing shoes shorter than their foot length (Menz and Morris 2005). This cramping of the toes in tight shoes over time, coupled with the ability of the foot to change structure, can promote the formation of toe deformities such as hallux valgus and claw and hammer toes. This is supported by study on a partial-shoe wearing community who found that hallux valgus was present in only 2% of non-shoe wearers (Shine 1965). In the same community where they had opted to wear shoes this percentage increased linearly with years spent wearing shoes and in the group who had worn shoes for over 60 years, 16% of men and 48% of women suffered from hallux valgus (Shine 1965).

Weakness within the foot may also play a role in contributing to foot problems. It is well understood that the intrinsic foot muscles contribute to maintaining foot posture (Fiolkowski, Brunt et al. 2003, Headlee, Leonard et al. 2008, Kelly, Cresswell et al. 2014). This has primarily been shown regarding the structural integrity of the longitudinal arch however a strong ABH may also play a role in preventing hallux valgus. The medial location of ABH with its function comprising hallux abduction and plantar flexion suggests a role in maintaining normal hallux position. In the aforementioned study by Aiyer et al. the size of ABH across age was assessed and a significant decrease in size was witnessed in older age hallux valgus patients (Aiyer, Stewart et al. 2015). Conversely in the study by Mickle et al. no decrease in size was witnessed in this muscle in older people without foot deformities (Mickle, Angin et al. 2016) possibly implying a protective role of ABH against hallux valgus.

Increased body weight also bears relevance to the prevalence of foot pain experienced in older age. In a population sample, 40% of obese individuals reported suffering from disabling foot pain compared to only 11% in non-overweight individuals (Mickle and Steele 2015). Additionally, although a causative relationship can’t be determined, obese individuals also had significantly reduced toe flexor strength suggesting impaired foot functionality (Mickle and Steele 2015). In a separate study it was also found that overweight males had an increased likelihood to suffer from hallux valgus (Nguyen, Hillstrom et al. 2010). Controlling bodyweight therefore seems pertinent in order to reduce the risk of developing disabling foot symptoms. Also of particular relevance is the reported link between foot problems and increased fall risk. Research studies have concluded a moderate relationship between foot problems and risk of experiencing a fall (Mickle, Munro et al. 2009)whilst suffering from multiple foot problems further increases this risk (Menz and Lord 2001). It is therefore essential to research and then implement ways to reduce the prevalence of foot problems in order to alleviate this increase in fall risk.

1.2.4 Age-Related Changes to Gait

Increased age is often accompanied by concurrent changes in gait performance and ability. The most commonly reported and consistent findings are that older age adults tend to walk slower and take shorter steps than their younger counterparts (Prince, Corriveau et al. 1997, McGibbon 2003). This reduction in gait velocity appears to be somewhat attributed to the age-related reduction in muscle strength and power; particularly in the ankle plantar flexors (Figure 3). The decline in force producing capacity of the ankle plantar flexors, which are thought to be responsible for body stabilisation and propelling the body forward into the swing phase of gait, results in shorter steps being taken and concurrently a lower gait velocity (McGibbon 2003). The reduction in ankle power may also have an effect on gait variability, and ultimately instability, with a positive correlation being reported between strength and stride to stride variability and higher variability values in fallers than non-fallers (Hausdorff, Rios et al. 2001).

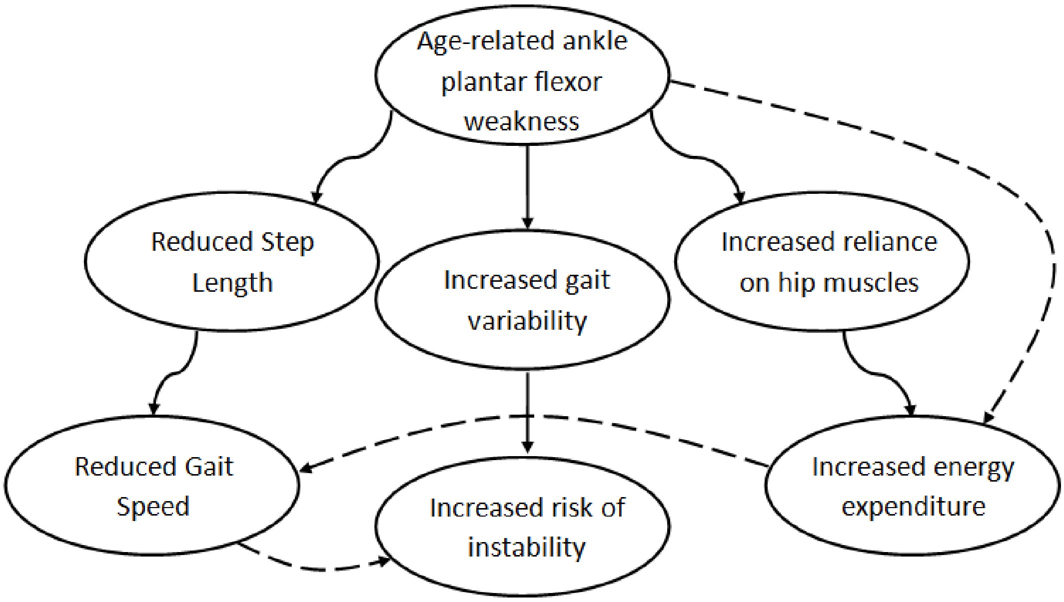

Figure 3 – An overview of the changes in gait performance which occur as a result of ageing and the observed associated decline in ankle plantar flexor force.

Further evidence of an impaired push off phase and initiation of swing through plantar flexor weakness in elderly gait is provided by foot pressure data. Elderly walkers had a significantly reduced pressure profile and lower mean force in the anterior region of the foot than the young group(Hessert, Vyas et al. 2005). Interestingly there was also an age effect witnessed at the heel strike and early stance phase (Figure 4). It was noticed that the elderly had significantly reduced mean pressure in the medial region of the foot during the gait phases and tended to preferentially weight bear on lateral region of foot (Hessert, Vyas et al. 2005). The authors highlight that this foot pressure distribution is similar to what is observed when the plantar surface of the foot has been experimentally desensitised in prior study (Eils, Nolte et al. 2002) and points to an age-related decline in proprioception and potentially at an increased risk of instability.

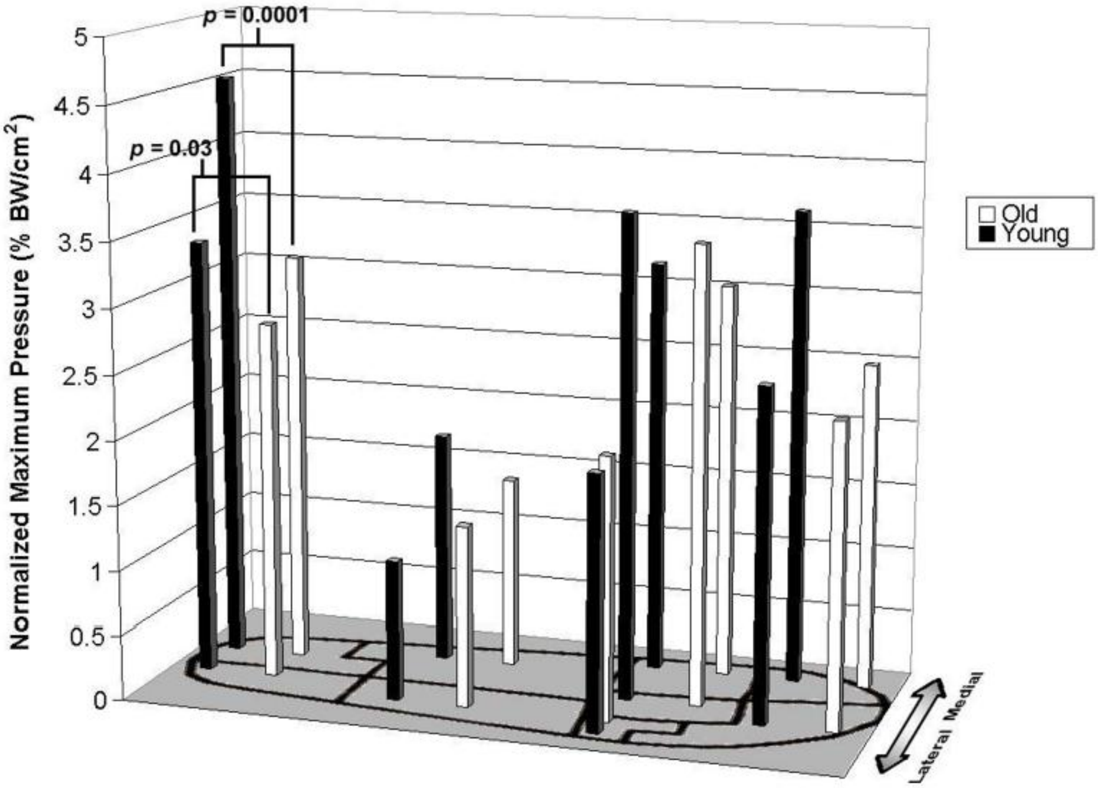

Figure 4 – The differences in maximum pressure distribution across the foot sole between young and old adults during treadmill walking. Heel is to the left. Figure taken from Hessert et al (2005) (Hessert, Vyas et al. 2005).

Alongside these spatial-temporal changes in gait, adaptations occur in the kinematics of elderly gait in order to compensate for the reduction in plantar flexor force output. More work is completed by the hip flexors in order to 'pull' the leg through into the swing phase to compensate for the lack of 'push' by the ankle plantar flexors whilst the hip extensors also increase their activation to stabilise the trunk (McGibbon 2003). In effect, neuromuscular adaptation occurs to preferentially distribute work to more proximal areas i.e. the hip, to compensate for the decline in ankle plantar flexor force (Figure 5).

A concomitant outcome which has also been demonstrated in the gait of older adults is that of an increased energy cost. Energy expenditure, as determined by oxygen uptake (ml.kg^{-1}.min^{-1}) has been shown to be significantly higher across 5 different walking speeds in older adults (mean age 81.6years) compared to young adults (mean age 24.6years)(Malatesta, Simar et al. 2003). Adults in the mid-old age group (mean age 65.3years) also had significantly higher energy expenditure than the young adults but only for the two fastest speeds (1.33 and 1.56m/s) implying a decline in economy with increasing age (Malatesta, Simar et al. 2003). This is supported by a prior study which reported an 8% increase in energy expenditure in older adults compared to younger adults during walking (Martin, Rothstein et al. 1992). It was suggested that the decrease in economy is due to the age-related reduction in muscular force requiring the muscle to work at a greater proportion of its total capacity as well as increased recruitment of other muscles (Martin, Rothstein et al. 1992). This can potentially further explain why older adults walk slower and also mean that they are less able to walk for long periods of time without suffering from fatigue.

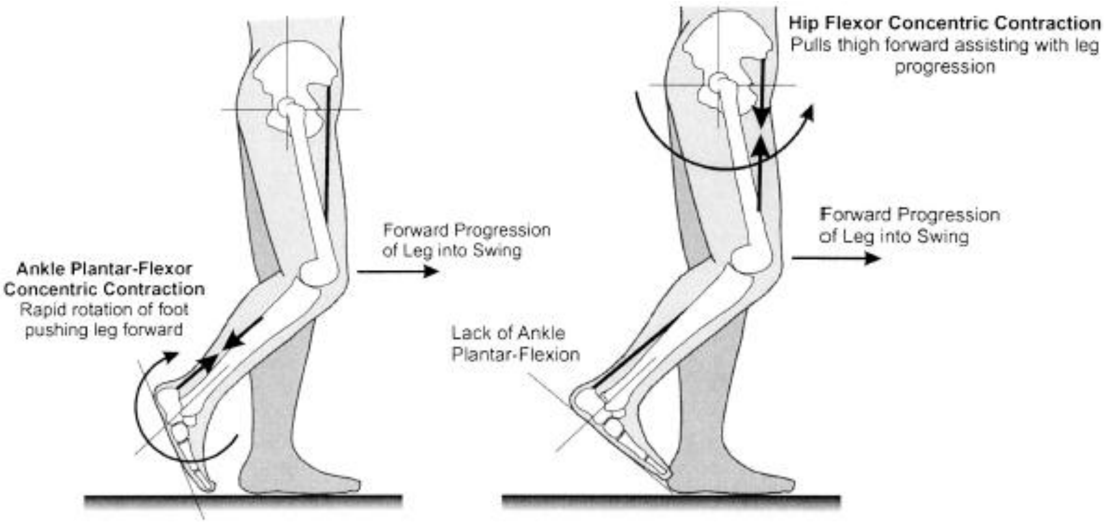

Figure 5 – The role of both the hip flexors and the ankle plantar flexors in propulsion and the initiation of the swing phase. Increased age tends to lead to the preferential increase in work by the hip flexors to account for the decline in ankle plantar flexor force. Figures taken from McGibbon et al. (2003) (McGibbon 2003).

An unfortunately common problem with ageing gait is the apparent inability to recover from a slip often resulting in a fall. It was initially reported that the reason for increased slip prevalence in older adults was due to a higher horizontal heel contact velocity coinciding with a reduction in hamstring activation and these changes would suggest an increased risk of slips due to a greater frictional force demand. However this doesn't seem to be the case as it has been shown that older adults do not have an increased required coefficient of friction and do not slip more regularly than younger adults but are simply unable to recover from the slip (Lockhart and Kim 2006, Moyer, Chambers et al. 2006). It is therefore in the recovery phase of a slip rather than the initiation where age differences are present. Research in this area indicates an inability to generate required joint moments at the ankle and knee due to muscle weakness (Liu and Lockhart 2009) alongside impaired compensatory postural reactions (Woollacott, Shumway-Cook et al. 1986) are the reasons for this reduction in balance recovery.

1.2.5 Falls: Prevalence, Cost and Consequences

A fall is defined by the World Health Organisation as ‘an event which results in a person coming to rest inadvertently on the ground or floor or other lower level’ and the aforementioned age-related changes lead to the older age population suffering from the greatest risk. It’s been reported that approximately 30% of people older than 65 and 50% of people older than 80 falling at least once a year (NICE 2013) and the Department of Health states that injuries caused by falls are the most common cause of death in people aged over 75 in the UK (Health 2010). A frequent consequence of falling, particularly in females, is suffering a hip fracture; falls are responsible for 90% of all hip fractures (Cummings and Melton 2002). In 2010, across the UK an estimated 79,000 hip fractures occurred with 56,000 of these in women (Svedbom, Hernlund et al. 2013). Suffering a hip fracture significantly reduces health-related quality of life (Adachi, Loannidis et al. 2001) and ~20% of previously community dwelling older adults require varying degrees of nursing care following fracture (Autier, Haentjens et al. 2000, Blume and Curtis 2011). As a result the cost per year per hip fracture patient has been estimated at approximately $40,000 in the United States of America (Blume and Curtis 2011) and in the UK £9,390 for the initial treatment plus £24,444 for the resulting yearly cost of a nursing home if required (Svedbom, Hernlund et al. 2013).

Fear of falling (FOF) is often a deleterious consequence of a prior fall and significantly lowers balance confidence (Lajoie and Gallagher 2004) which itself has been shown to be a major independent predictor of falling (Landers, Oscar et al. 2016). FOF and low balance confidence are also significantly related to a reduction in physical activity and social isolation which puts a restriction on life-space mobility and ultimately can lead to frailty and mortality (Auais, Alvarado et al. 2017). As such there are many studies aimed at investigating ways to improve balance confidence to prevent these adverse consequences and these have been summarised in a recent review. The interventions ranged from various exercise programmes (such as resistance training, agility training, functional balance training and Tai Chi), multi-factorial treatment (including exercise with medication adjustments and behavioural instructions, occupational therapist home visits to decrease home hazards/home rehabilitation programme upon discharge), psychological interventions (mental imagery) and additional aids (hip protectors, personal emergency response system). Significant improvements in balance confidence were found for the exercise programmes with Tai Chi being the most effective in addressing low balance confidence (Rand, Miller et al. 2011). The authors explain that Tai Chi may be the most effective at improving balance confidence due to it addressing both the sensory motor aspects of balance as well as cognitively through promoting emphasis on relaxation, awareness and focus (Rand, Miller et al. 2011). The link with relaxation is key as FOF has been shown to lead to a 'stiffening' strategy which compromises performance in dynamic tasks and ultimately leads to an increased risk of falling (Young and Mark Williams 2015). The authors indicate that stiffening behaviours lead to impaired sensory information via an internal focus of attention which negatively affects the planning and execution of movements(Young and Mark Williams 2015). It is apparent that improving balance confidence is a vital factor in reducing the risk of falls and associated detriments in health and wellbeing.

The beneficial effect of exercise on balance confidence also corresponds to the fall itself. Research shows that there are significantly fewer falls in older age in those people who remain sufficiently active throughout their life (Wright, Robinson et al. 2012). Additionally, remaining fit and active throughout life can have a protective effect even if a fall is experienced. Previous study has shown that regular weight-bearing or resistance exercise, reduces the risk of osteoporosis and fall-related fractures through increasing bone mineral density and reducing age-related bone loss (Miller, McClave et al. 2016). This protective effect of exercise on both fall risk and fall fracture incidence clearly demonstrates how emphasis needs to be placed on ensuring a good level of physical activity is maintained throughout life and especially in older age.

Footwear choices are also a factor in decreasing or increasing the risk of falls in older age and this will be discussed in the following section.

1.3 Footwear

1.3.1 The Evolution of Footwear

In comparison to how long our common hominin ancestors have been suggested to be walking bipedally (~6-7million years) (Crompton, Vereecke et al. 2008); footwear is perceived to be a fairly new addition. It is difficult to pinpoint exactly when wearing footwear became frequent but some research suggests it is around the middle Upper Paleolithic era (~30,000 years ago). This is based on anatomical archaeological data on the toes of a middle Upper Paleolithic sample which implies the introduction of an external matter around the foot (Trinkaus 2005). This sample displayed a reduction in the size of the three middle lesser toes, as well as length of the hallux, compared to middle Paleolithic samples but no difference in femoral or tibial robusticity (Trinkaus 2005). This suggests that an external matter may have been introduced around the foot at this time which had the effect of limiting the habitual loads and reliance on the lesser toes during walking resulting in decreased robusticity in these areas(Trinkaus 2005). Barefoot footprints have also been found in caves of similar date origin and thus it appears that footwear use was likely limited to times when protection of the foot was deemed necessary rather than at all times (Trinkaus 2005).

With the primary purpose of prehistorical footwear likely being protection from abrasions and cold weather their design was simple but effective. One of the earliest resemblance of intact shoes to be uncovered was found within the Arnold Research Cave in Missouri and dates back to ~8000 years ago (Kuttruff, DeHart et al. 1998). They were made out of a fibrous plant material and of sandal design such that they covered the soles of the feet and had a tie system over the foot which attached around the ankle (Kuttruff, DeHart et al. 1998). More recently there was a discovery of the oldest leather shoe on record which was found in a cave in Armenia(Pinhasi, Gasparian et al. 2010). This leather shoe was dated back to ~3500 years ago suggesting that shoe design had progressed from the use of fibrous plant material to animal hide material (however the difference in location of the discoveries may also be a factor in this) (Pinhasi, Gasparian et al. 2010). The shoe design was different to the sandals previously

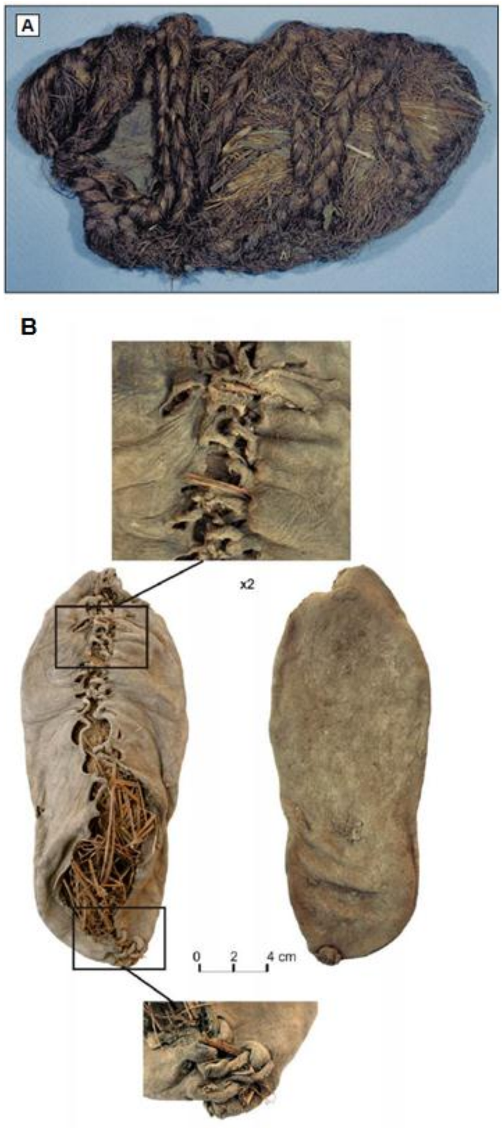

Figure 6 – Top (A): an early sandal made from fibrous plant material which has been radiocarbon dated back to ~8000years ago. Taken from Kuttruff et al (1998) (Kuttruff, DeHart et al. 1998).
Bottom (B): one of the earliest leather shoes to be uncovered and has been dated back to ~3500 years ago. Taken from Pinhasi et al (2010) (Pinhasi, Gasparian et al. 2010)

found in that it was made out of a single piece of leather wrapped around the foot and sewn along the dorsal portion to fully encompass the foot. The shoe was found stuffed with grass which it was suggested to be for extra warmth or to help maintain shape (Pinhasi, Gasparian et al. 2010). Other leather footwear were also discovered in the Arnold Research Cave discovery which dated back ~1000 years but were shown to be of a moccasin style design using leather to form sides, back and a rounded toe box with again grass lined within the shoe (Kuttruff, DeHart et al. 1998). This appears to demonstrate how footwear design progressed over time, using new materials, adapting the style and introducing innovations for comfort and warmth. It could have been that even thousands of years ago footwear was used as a way to impart artistic creativity, much like the making and wearing of jewellery or to demonstrate status.

In Roman times, footwear played a major role in terms of developing the Roman civilisation. The Roman sandal, or caliga, with the hobnailed sole provided the soldier with a durable and protective shoe which allowed them march on rough terrain and travel further than other armies (van Driel-Murray 2001). As the Romans travelled they took on styles from other civilisations such as the gallica which was a wooden-soled boot worn by the Gaul warriors in France. The footwear worn was a symbol of power to the Romans and officers wore a boot to designate rank and these were decorated and worn higher up the leg depending on the rank of the officer (van Driel-Murray 2001).

Britain began producing its own shoes once the Romans left and as time passed it seems clear that footwear became less about protection from the environment and more about a medium for artistic creativity or to demonstrate status (Riello and McNeil 2006). During the Medieval age the length of the toes became a symbol of power and status and hence the King wore shoes with the longest toes whilst leather ankle shoes were popular around this time for the majority (Staheli 1991). Across Europe heavy wooden shoes became popular in the form of clogs or sabots which were often made for lower class labour workers. Meanwhile in China, dating back to the 10th or 11th century, women were being subjected to foot binding; a practice of constricting the feet of young girls to modify the shape and keep them small (Riello and McNeil 2006). It is thought it was done as a symbol of status such that if women do not have to use their feet to work they are of a higher social standing but it was later adopted as a sign of beauty and persisted for centuries until being officially banned in 1912.

In Europe up until around the 16th century most footwear had been made with flat soles and it wasn't until this time that heels began to emerge(Staheli 1991). Wearing heels was understood to portray wealth and social standing, with royalty wearing the highest heels, and this is the reason for the term 'well-heeled' (Staheli 1991). However, adding a heel to the shoe made the production of footwear more difficult and therefore during this period shoes were often made straight such that they could fit on either foot which made production easier but comprised comfort. It wasn't until the mid 1800s when shoes began to be made specifically for the right and left feet again when a decline in heel height occurred leading to the introduction of pump and dress shoes.

In the late 19th century rubber began to be used in shoe making to create the soles as an alternative to leather and the first shoe to be created this way was the plimsoll (Riello and McNeil 2006). These shoes consisted of a rubber sole which was glued to a canvas upper and paved the way for the design of future lightweight canvas athletic shoes. In the early 20th century the first mass market sneaker or trainer, called Keds, was produced and at around the same time Converse released their first trainer, the Converse All Stars, aimed at basketball players. This started the trend of specialist companies, such as Adidas, Puma and later, Nike, creating footwear specific for different sporting events throughout the mid 20th century. Due to its lightweight and comfortable design coupled with the opportunity for inventive freedom this type of footwear soon became the footwear choice of everyday life and not just for sporting activities.

In the late 20th century and up to the modern day, footwear has become increasingly complex and rather than comprising one piece of material wrapped around the foot as in the first leather shoe from ~3500 years ago, now contains numerous individual features which are then assembled. As displayed in Figure 7, most modern shoes generally have 7 main features; the outsole, the midsole, the shank, the insole, the heel counter, the toe box and the upper. The outsole is the bottom portion of the shoe which is contact with the ground and is typically made of rubber. It usually also has lugs or some degree of texture built in for extra traction. Above the outsole and normally glued to it is the midsole. The midsole is often made out of EVA foam and its main purpose is to provide shock absorption or motion control and this is dependent on the density of the foam used. Softer density will absorb more shock whilst harder density will compress less and prevent movement. These can often be combined across various areas of the midsole to provide shock absorption in some areas whilst controlling motion in others and this is termed a ‘dual density’ midsole. Encompassed within the midsole is the shank which is a rigid structure providing support along the longitudinal aspect of the shoe between the heel and the mid-tarsal joint and again this is a motion control feature. Above the midsole is the insole which is the layer of material which fits atop the shoe sole. It is often made out of foam to add extra comfort or support to the wearer. At the rear of the shoe is the heel counter. This is made out of a stiff material and provides medial and lateral support whilst strengthening the rear of the shoe to help retain its shape. It should grip against the rear of the foot and surround the Achilles tendon. The toe box is the front portion of the shoe where the toes are situated. It retains the shape of the forefoot based on the shoe last; the structure used to mould the shape and design of the shoe, and it should be wide enough and high enough to allow room for the toes and have a degree of rigidity to protect the toes. Both the toe box and heel counter form part of the upper which is the portion of the shoe which covers the dorsum and attaches to the midsole. This can be made out of numerous materials both natural and synthetic which are aimed at being breathable to allow for airflow and temperature regulation and often offer a degree of waterproofing. The upper also incorporates lacing or buckles to give the wearer a degree of adjustment to improve the fit around the foot.

Figure 7 – Displaying the 7 main features of a modern day shoe.

Footwear production is now a huge business and it continues to grow with reports that 22 billion pairs of shoes were made worldwide in 2013 which shows an increase of 1 billion compared to the relative figures from 2011 (APICCAPS 2012, APICCAPS 2014). Regardless of the capital gains associated with the industry, what needs to be ensured is that footwear remains functional as per its original purpose of providing protection and warmth and does not allow external pressures of fashion and aesthetics to contribute to a decline in foot health through dysfunctional design.

1.3.2 Footwear Effects on Foot Structure

There is evidence to suggest that the long term wearing of modern day constrictive footwear may lead to changes in foot shape, structure and function. Recently a comparative study between habitual unshod and shod populations was completed to determine differences in foot morphology (Shu, Mei et al. 2015). Using a 3D foot scanner it was found that the females of the unshod group had significantly wider feet, reduced inward deviation of the hallux and a greater distance between the hallux and second toe (Shu, Mei et al. 2015). As this was present in only the females it further suggests that footwear may be the main reason for these observed differences due to the propensity of females to wear ill-fitting, narrow closed toe shoes more so than males (Menz and Morris 2005). Furthermore, a previous study between a habitually shod population and a population who had never worn closed toe shoes (but did wear sandals or flip flops) demonstrated that the shod group had a considerably reduced pliability indicating increased stiffness and reduced mobility (Kadambande, Khurana et al. 2006). The authors suggest that the narrow nature of modern day toe boxes causes a reduction in the ability of the forefoot to spread as it would unshod and thus leading to stiffer less mobile feet. They also state that an inbuilt medial arch support common in modern day footwear may also restrict the compressive action of the longitudinal arch of the foot during weight bearing conditions. It is worth noting however that the populations may have inherently different foot characteristics due to their different ethnic backgrounds; the sample groups who were habitually unshod/had never worn closed toe shoes were Indians whilst the habitual shod sample groups were British (Kadambande, Khurana et al. 2006) or Chinese(Shu, Mei et al. 2015).

To further this research question another study has investigated the difference in foot shape between habitually shod (Western Caucasian in Belgium) and a habitual unshod group (native Indians) whom in this case had never worn shoes and in extremely rare cases flip flops (D’AoÛt, Pataky et al. 2009). Interestingly this study also included a third group of the same native background as the unshod group but who now wore shoes (mostly flip-flops or sandals) on a daily basis after walking barefoot mostly as a child (D'AoÛt, Pataky et al. 2009). Aside from the pure comparison between Indian barefoot and Western Caucasian shod groups this also enabled a comparison between the similar native Indian groups to examine if the introduction of footwear after childhood had provoked any changes in foot shape or structure. Clear differences in foot shape were witnessed with the Western shod group displaying considerably shorter foot lengths with respect to stature, considerably narrower feet with respect to foot length and consequently a smaller foot area (D'AoÛt, Pataky et al. 2009). Interestingly the shod Indian group were observed to be intermediate in the foot width measurement potentially suggesting that the introduction of a boundary around the foot in the form of sandals may have begun to have an impact on the structure of the foot and its ability to spread under load.

This study also explored how the plantar pressure profile during walking differed across these three populations (D'AoÛt, Pataky et al. 2009). The findings suggest that the difference in foot shape may also lead to a change in the distribution of pressure across the foot during the weight bearing phase of locomotion. It was observed that the Western shod group had distinctly higher localised peak pressures and pressure impulses at the heel and 2nd and 3rd metatarsal regions when compared to the barefoot and shod Indian groups (D'AoÛt, Pataky et al. 2009). Both Indian groups had greater relative plantar pressure values across the midfoot and toe region and a greater spread of load across the metatarsal and heel regions compared to the Western group. This is likely due to the wider foot shape and improved distribution of load across the foot but also potentially greater pliability and arch compressive function indicative of the increase in pressure in the midfoot region in these groups. With the shod Indian group observed to exhibit intermediate peak plantar pressure values with respect to the other two groups it suggests that the introduction and use of footwear may have an impact on both the foot shape and the associated plantar pressures experienced during walking (D'AoÛt, Pataky et al. 2009).

Further study has gone on to analyse the impact of 3 common toe box shapes, namely round, square and pointed, on the dorsal and plantar pressures across the foot (Branthwaite, Chockalingam et al. 2013). Previous study has highlighted how often the footwear worn, particularly in females, is too narrow for the foot shape with data suggesting two thirds of elderly females are wearing shoes which have a toe box too narrow to accommodate the width of their forefoot (Chantelau and Gede 2002, Menz and Morris 2005). This narrow footwear can lead to high pressures being experienced on the forefoot resulting in associated foot pain and potentially the increased prevalence of foot deformities and calluses (Paiva de Castro, Rebelatto et al. 2010, Branthwaite, Chockalingam et al. 2013). This suggestion is supported by the fact that foot deformities such as hallux valgus are increasingly common within the Western shod community but substantially less so in unshod or partially shoe wearing communities (Shine 1965, Kadambande, Khurana et al. 2006).

1.3.3 Footwear Effects on Foot Function

Footwear has been suggested to somewhat constrain natural foot motion during walking or running and as a result potentially impairs its function. Using a multi-segment foot model it was shown that when walking barefoot there was greater torsional and adduction range of motion of the foot (Morio, Lake et al. 2009). Furthermore, during late stance the rate of eversion was reduced considerably when shod compared to barefoot suggesting that the constrains of the footwear not only limit the range of motion but also modified the pattern of motion during the push off phase (Morio, Lake et al. 2009).

Previous research has suggested that certain aspects of footwear may interfere with the sensorimotor system reducing its sensitivity and control. In one study the midsole, the footwear feature responsible for providing support and cushioning, was modified in terms of thickness and hardness and participants were required to walk along a balance beam barefoot and in six different midsole variations (Robbins, Waked et al. 1994). It was observed that participants suffered more falls from the beam, termed balance failures, when wearing shoes with thick soft midsoles, deemed the most cushioned and supportive, whereas had improved stability in shoes with thin hard midsole properties (Robbins, Waked et al. 1994). It's suggested that thin hard soled shoes are optimum for older adults as they offer reduced interference to foot position awareness and correspondingly improve stability (Robbins, Waked et al. 1997). A similar finding was found in a static assessment of foot position awareness(Robbins, Waked et al. 1995). Participants were required to stand on a sloped surface and estimate the amplitude and direction of the surface slope when barefoot and in a pair of common athletic shoes(Robbins, Waked et al. 1995). Their vision of the surface was removed so they had to rely on the proprioceptive feedback from the ankles to predict the slope. It was found that participants were better at predicting the slope of the surface when barefoot as opposed to when wearing the athletic footwear suggesting an impairment of proprioceptive sensibility when wearing this type of footwear (Robbins, Waked et al. 1995). It is proposed that the compression of the shoe sole material underfoot may provide an erroneous input to the sensorimotor system, particularly mechanoreceptors, regarding foot and leg position resulting in the decline in awareness and stability (Robbins, Waked et al. 1997).

One potential benefit of being barefoot to the sensorimotor system has been put forward in a recent position paper. It has been suggested that being barefoot activates all of the muscles in the foot-ankle complex including the smaller muscles and that these muscles are better situated to detect smaller changes in foot position more quickly and with less force being required to correct (Nigg 2009). Through increasing the activation of these muscles with 'barefoot training' the author suggests that this could promote increased joint stability (Nigg 2009). Through the same mechanism it may also be the case that overall body stability could be enhanced through the improvement in sensorimotor sensibility as previously mentioned.

1.3.4 Minimalist Footwear

Alongside modern day footwear with built in cushioned and supportive features have emerged minimalist footwear. A consensus definition explaining the main features of these shoes that differentiate them from conventional footwear has recently been suggested and is as follows:

"Footwear providing minimal interference with the natural movement of the foot due to its high flexibility, low heel to toe drop, weight and stack height, and the absence of motion control and stability devices" (Esculier, Dubois et al. 2015).

The concept of this footwear, as the definition suggests, is to limit the interference with the function of the foot experienced with conventional footwear by solely offering a protective surface from abrasions. In other words the foot is placed back into a natural position of rearfoot and forefoot vertical alignment, there are no arch supports or midsole cushioning/stability structures returning this responsibility back to the foot itself and the sole is thin and flexible rather than thick and rigid to reduce attenuation of plantar surface sensory information via material interference underfoot. There has been an increase in the research investigating this type of footwear in recent years but primarily in terms of running and improving performance/reducing injury incidence. Whilst these research findings cannot be entirely applied to walking and the potential benefits to everyday life or the older age population, they do offer a basis and primary evidence for the previously proposed benefits of this footwear.

In relation to the previous discussion on foot position awareness impairment with athletic footwear, more recent research has explored how static and dynamic position sense is affected by minimalist footwear (Squadrone and Gallozzi 2011). Static position sense was examined via a similar slope estimation task as used previously (Robbins, Waked et al. 1995) whilst dynamic position sense was determined via running on a treadmill with varying degrees of inclination. Both tests required the participant to estimate the inclination (following reference slope angles) when barefoot, in a minimalist shoe and in a conventional running shoe. It was found that static position sense was considerably worse in the conventional running shoe compared to barefoot and the minimalist shoe whilst the minimalist shoe also displayed considerably better position sense in the dynamic task (Squadrone and Gallozzi 2011). This study appears to suggest that the reduction in material composition between the foot and the ground does serve to improve foot position awareness in both static and dynamic conditions most likely through muscle mechanoreceptors rather than cutaneous.

Similarly several recent studies have supported the notion that removing the supportive/cushioned structures leads to increased activation of the intrinsic foot muscles (Miller, Whitcome et al. 2014, Chen, Sze et al. 2016, Johnson, Myrer et al. 2016). A 12-week training study demonstrated that running in minimalist shoes resulted in greater cross-sectional area (CSA) and muscle volume (MV) increases in the FDB muscle than running in conventional footwear (Miller, Whitcome et al. 2014). Furthermore CSA and MV of the abductor digit minimi as well as longitudinal arch stiffness also increased significantly in the minimalist shoe group only (Miller, Whitcome et al. 2014). In support, following transitioning from conventionally shod running to minimally shod running over a 6 month period, significant increases in leg and foot muscle volume were witnessed whilst the control group who remained wearing conventional running shoes showed no changes (Chen, Sze et al. 2016). It therefore appears clear that the increased loading is placed on lower leg and foot musculature following the removal of cushioning and stability structures and the increases in strength witnessed in these studies are a testimony to this. Reciprocally it can be implied that wearing footwear with high stability and cushioning properties reduces the reliance on this musculature and hence may lead to potentially detrimental weakening over time. It therefore should be stated that care should be taken when transitioning to minimal footwear, primarily for running due to the high loads experienced, as sufficient strength in these muscles is required to cope with the increased demand. This is highlighted by the presence of bone marrow oedema witnessed during a transition to minimalist footwear over 10 weeks in runners with significantly smaller muscle size (Johnson, Myrer et al. 2016).

Furthermore, the premise of maintaining the natural movement of the foot is also supported by previous research findings. In a study on children it was observed that many foot motion variables including hallux flexion and forefoot width were reduced when walking in conventional shoes compared to the barefoot condition however when walking in more flexible footwear the differences to barefoot were reduced (Wolf, Simon et al. 2008). It therefore appears that minimalist footwear may be effective at allowing for more natural foot motion than conventional footwear but still constrains to some extent. The research on foot motion when wearing minimalist footwear during walking however is sparse and conclusions cannot be made until more research is completed.

1.3.5 Footwear Effects on Gait

The difference between walking barefoot and walking in footwear has been explained extensively through the completion of a systematic review on this topic. This is located in Chapter 2 where this topic is explored and summarised.

1.4 Thesis Aims and Objectives

The overall aim of this thesis is to investigate if minimalist footwear may be beneficial in terms of improving the gait and balance ability of older adults. The thesis is split into chapters representing individual studies focussing on specific research questions regarding minimalist footwear. Two chapters have already been published (Chapter 2 – **Franklin S, Grey M.J, Heneghan N, Bowen L, Li F.X.** ***Barefoot vs. common footwear: A systematic review of the kinematic, kinetic and muscle activity differences during walking,*** **Gait & Posture, 2015,** 42**(3), 230-239**); (Chapter 4 **– Franklin S, Li F.X, Grey M.J.** ***Modifications in lower leg muscle activation***

when walking barefoot or in minimalist shoes across different age-groups. **Gait & Posture, 2017, 60, 1-5**). The aims of these chapters are summarised below:

- Chapter 2 – A systematic review of the current literature on the kinematic, kinetic and muscle activity differences when walking barefoot or in common footwear. This chapter aims to provide a clear summary of what is already known about how walking barefoot differs from walking in common footwear and also how minimalist footwear may fit into the equation.
- Chapter 3 – This chapter serves to fill in the lack of research on minimalist footwear highlighted from the systematic review. It investigates if walking in minimalist footwear is comparable to walking barefoot in terms of gait kinematics and kinetics. It also examines if different age groups respond differently to walking barefoot or in minimalist footwear as a result of years spent wearing conventional footwear.
- Chapter 4 – This chapter also serves to fill in gaps in the research highlighted from the systematic review. It explores if walking barefoot or in minimalist footwear increases the muscle activation of lower leg muscles in comparison to walking in conventional footwear. It also investigates this with respect to age and years spent walking in conventional footwear.
- Chapter 5 – This chapter is a methodological chapter whereby a custom built device was tested for its reliability at measuring hallux plantar flexor force. This device was then used in the following experiments.
- Chapter 6 – This chapter investigates if a 6-week home based foot exercise program would be effective at increasing the foot muscle strength of older adults. It also aims to examine if increasing foot muscle strength leads to improvements in balance and gait performance of older adults.

- Chapter 7 – This chapter explores if wearing minimalist footwear in daily life for a 4 month period can be a method in which to improve the foot muscle strength, balance and gait stability of older adults. It aims to examine if increased muscle activation is promoted when standing and walking in minimalist footwear in comparison to walking in conventional footwear consequently leading to improvements in foot muscle strength. It also aims to determine if pertinent improvements in proprioceptive sensibility are evoked following extended use of minimalist footwear leading to improvements in balance and gait stability.
- Chapter 8 – This chapter presents a general discussion of the major findings from the aforementioned experimental chapters, an overall summary of the outcomes from this thesis and suggestions of how future research should progress in this area.

Published: Gait & Posture, 42, (2015), 230-239

Chapter 2

Barefoot vs. common footwear: A systematic review of the kinematic, kinetic and muscle activity differences during walking

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Abstract

Habitual footwear use has been reported to influence foot structure with an acute exposure being shown to alter foot position and mechanics. The foot is highly specialised thus these changes in structure/ position could influence functionality. This review aims to investigate the effect of footwear on gait, specifically focussing on studies that have assessed kinematics, kinetics and muscle activity between walking barefoot and in common footwear. In line with PRISMA and published guidelines, a literature search was completed across six databases comprising Medline, EMBASE, Scopus, AMED, Cochrane Library and Web of Science. Fifteen of 466 articles met the predetermined inclusion criteria and were included in the review. All articles were assessed for methodological quality using a modified assessment tool based on the STROBE statement for reporting observational studies and the CASP appraisal tool. Walking barefoot enables increased forefoot spreading under load and habitual barefoot walkers have anatomically wider feet. Spatial-temporal differences including, reduced step/stride length and increased cadence, are observed when barefoot. Flatter foot placement, increased knee flexion and a reduced peak vertical ground reaction force at initial contact are also reported. Habitual barefoot walkers exhibit lower peak plantar pressures and pressure impulses, whereas peak plantar pressures are increased in the habitually shod wearer walking barefoot. Footwear particularly affects the kinematics and kinetics of gait acutely and chronically. Little research has

been completed in older age populations (50+ years) and thus further research is required to better understand the effect of footwear on walking across the lifespan.

Introduction

Humans are one of the few species who have mastered bipedal locomotion and their foot has evolved to be the basis for such a specialised gait. The human foot alone comprises 26 bones, 33 joints and 19 muscles (Theodore Dimon 2008). The bones are arranged to form a medial longitudinal arch which makes it ideal for its function of supporting the weight of the body and spreading the forces experienced during gait (McKeon, Hertel et al. 2015). Aside from the structure of the bones there is a complex array of muscles, both internal and external of the foot, which combine with the somesthetic system to control balance and movement (Kavounoudias, Roll et al. 2001). Kennedy et al (Kennedy and Inglis 2002) reported the presence of 104 cutaneous mechanoreceptors located in the foot sole. Furthermore receptor distribution was primarily where the foot is in contact with the ground, and when the foot was unloaded no background activity was found. In addition there are more fast adapting units than slow suggesting a high dynamic sensitivity (Kennedy and Inglis 2002). Collectively these factors evidence the role of the human foot in balance and movement control but what is less clear is the impact of wearing shoes on the human foot and whether this may influence movement control and associated variables during walking gait.

Anthropological evidence suggests that footwear began to be worn approximately 40,000 years ago (Trinkaus and Shang 2008). This is hypothesised based on the observations of a reduction in toe length at this time indicating a reduction in reliance on and loading of the lesser toes during locomotion (Trinkaus 2005). Furthermore as footwear has evolved from simple open-toe sandals to more complex items of fashion, with their design being increasingly dependent on aesthetics, the potential impact on foot function has been overlooked. Pointed toe and closed toe shoes have become increasingly prominent in Western societies and the restriction of area within the toe box potentially contributes to, now deemed common, toe deformities such as hallux valgus, a valgus deformity on the first metatarsophalangeal joint (Al-Abdulwahab and Al-Dosry RD. 2000). This is particularly a problem in advanced age with the over two thirds of the older population's feet being considerably wider than the footwear available (Chantelau and Gede 2002). Additionally research has reported that wearing high heels of 5cm or higher over a minimum of a two-year period has significant effects to the muscle-tendon unit at the ankle (Csapo, Maganaris et al. 2010, Cronin, Barrett et al. 2012). Csapo and colleagues (Csapo, Maganaris et al. 2010) found a significant reduction in the gastrocnemius medialis fascicle lengths and significantly greater Achilles' tendon stiffness in the high heels group, resulting in a more plantar flexed ankle position at rest and a reduced active range of motion. This demonstrates the modifiable nature of the foot-ankle complex and the importance of wearing appropriate footwear to maintain good foot health and function.

Research has also shown how certain footwear can directly influence function. A common feature of modern athletic footwear is that of increased sole thickness which is marketed as providing cushioning against harmful impacts. Recent research has demonstrated that wearing this type footwear evokes significantly increased activation in the Peroneus Longus suggesting greater interference to ankle stability (Ramanathan, Parish et al. 2011). Moreover, footwear has been shown to hinder the kinesthesia (Robbins, Waked et al. 1995), with greater awareness of foot position observed in volunteers standing barefoot compared with wearing athletic footwear. Whilst these studies are limited to investigation in standing, the findings suggest the possibility that footwear could be interfering with the functional ability of the human foot and if this corresponds to changes in gait.

The aim of this review is to systematically review the research investigating kinematic, kinetic and muscle activity variables during walking barefoot and in normal footwear to help improve our understanding of how footwear influences gait pattern.

Methods

Study design and Search Strategy

Reporting in line with PRISMA guidelines (www.prisma-statement.org) and through consultation with subject specific and systematic review experts the literature review methodology was developed. The literature search was performed across a variety of databases (Medline, EMBASE, Web of Science, Cochrane Library, SCOPUS and AMED) encompassing publications within the years of 1980-January 2014. The search strategy employed across the electronic databases is presented below:

1. barefoot
2. walk*
3. exp Gait/
4. exp Locomotion/
5. kinematic*
6. kinetic*
7. exp Electromyography/
8. EMG
9. muscle activ*
10. 7 or 8 or 9
11. 5 or 6 or 10
12. 2 or 3 or 4
13. 1 and 12
14. 11 and 13

Study Selection

One reviewer (SF), who had received training on database searching, completed all searches which were independently checked by a second author (LB). Differences of opinion were resolved through discussion or a third author. Citation checking and search of grey literature, including key conference proceedings within the last 3 years was also undertaken. Authors were subsequently contacted to determine if any relevant proceedings had since reached publication.

Inclusion criteria were determined a priori. Studies were required to assess gait characteristics between footwear in terms of spatial-temporal variables, kinematics, kinetics, and muscle activity and behaviour. Participants were to be healthy and able to ambulate independently such that their gait pattern was considered normal and would not influence comparisons between footwear conditions. They could be of any age group and either gender to observe any differences throughout age and include data from both males and females to draw comparisons from if possible. Overground walking and treadmill walking were both deemed acceptable in order to access all studies analysing barefoot walking gait characteristics. Studies of observational cross-sectional design were included to allow for review of the comparison between footwear conditions wear inclusive of socks, open-toe footwear such as sandals or flip-flops and slippers. Observational comparative studies were deemed suitable if they were comparing between habitually barefoot, who have grown up and continue to live without wearing shoes, and habitually shod, who wear shoes on a day-to-day basis, populations to determine changes which occur over long term use with or without shoes. Case-control studies were also included providing the control group fitted the participant criteria and data was available for conclusions to be drawn solely from this group with regard to footwear intervention. If both groups fitted the participant criteria, then providing that data was available these were included and comparisons were focussed on the separate group's response to the footwear intervention rather than the comparisons between groups.

Studies were excluded if the footwear included any interventions aside from the features included in the original footwear design such as separate insoles or orthotics. Any studies involving participants who required a walking aid to ambulate were also excluded along with participants who had a known previous or current gait disorder or condition that could influence their gait (unless the study also consists of a control group through which analysis can be drawn from). Studies were excluded if they used running, unless a walking test was also completed from which analysis could be solely focussed. Literature other than peer-reviewed journal articles and comparative studies were excluded from the review.

Data collection and items

Using a standardised form the lead reviewer independently extracted the data. Study characteristics included repeated measures designs between various footwear conditions and between subject comparisons in terms of habitual barefoot and habitual shod users. Included outcomes were any measures which assessed spatial-temporal, kinematic, kinetic or muscle activity/behaviour variables.

Risk of bias across studies

To assess the methodological quality a bespoke critical appraisal tool was developed based upon the STROBE Checklist (Kavounoudias, Roll et al. 2001) for reporting observational studies and the CASP appraisal tool(Theodore Dimon 2008). All articles were assessed on these questions which determine if all the required steps for successful scientific reporting were taken and if the relevant information is presented clearly in the scientific paper. A score of 1 was given for each question if the article satisfied the question and a 0 given if it failed to do so. A total score out of 20 was then given for each paper. The quality assessment of the selected studies was carried out by one reviewer (SF) and then repeated independently by a second author (LB). Any issues were discussed to achieve consensus of opinion.

Synthesis of results

It was not appropriate to combine studies for meta-analysis, therefore the results were tabulated for semi quantitative comparison of spatial-temporal, kinematic, kinetic and muscle activity/behaviour variables.

Results

Search Results

The database search was completed in January 2014 and resulted in 924 records (155 Medline, 236 EMBASE, 222 Web of Science, 58 AMED, 9 The Cochrane Library, 244 SCOPUS) and a further 7 records were included from hand searches of reference lists, conference proceedings and contacting relevant authors in the field. Following removal of duplicates there were 466 records remaining from which analysis of titles and abstracts was undertaken. Twenty one articles were selected for full text screening of which 15 were deemed to meet the inclusion criteria and were subsequently used in the analysis. See Figure 8.

Methodological Quality

In five articles (Oeffinger, Brauch et al. 1999, Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Sacco, Akashi et al. 2010, Cronin and Finni 2013) there was no description of the footwear characteristics and/or type of footwear worn in the trials. Seven articles (Carl and Barrett 2008, Wolf, Simon et al. 2008, Morio, Lake et al. 2009, Keenan, Franz et al. 2011, Scott, Murley et al. 2012, Chard, Greene et al. 2013, Zhang, Paquette et al. 2013) used a standardised shoe across participants or controlled for the type of footwear worn. Of the fourteen articles which consisted of footwear trial conditions eight (Oeffinger, Brauch et al. 1999, Keenan, Franz et al. 2011, Wirth, Hauser et al. 2011, Scott, Murley et al. 2012, Chard, Greene et al. 2013, Cronin and Finni 2013, Tsai and Lin 2013, Zhang, Paquette et al. 2013) were administered in a random order to avoid carry over effects. Only one study reported details on the period of recruitment, exposure and data collection as well as the setting and location (Lythgo, Wilson et al. 2009). Four studies failed to report any demographic information of their participants (Oeffinger, Brauch et al. 1999, Carl and Barrett 2008, Wolf, Simon et al. 2008, Lythgo, Wilson et al. 2009). Only two studies reported how they derived their sample size (Chard, Greene et al. 2013, Zhang, Paquette et al. 2013) whilst only 3 studies reported effect sizes to illustrate the magnitude of the effect (D'AoÛt, Pataky et al. 2009, Sacco, Akashi et al. 2010, Scott, Murley et al. 2012). The breakdown of the results of the critical appraisal for each article is displayed in Table 1.

Study characteristics

The included studies were conducted in a variety of areas. These included Australia (Lythgo, Wilson et al. 2009, Scott, Murley et al. 2012, Chard, Greene et al. 2013), the USA(Oeffinger, Brauch et al. 1999, Carl and Barrett 2008, Keenan, Franz et al. 2011, Zhang, Paquette et al. 2013), Taiwan(Tsai and Lin 2013), Germany(Wolf, Simon et al. 2008), France(Morio, Lake et al. 2009), Brazil(Sacco, Akashi et al. 2010), Switzerland(Wirth, Hauser et al. 2011), Finland(Cronin and Finni 2013), Mexico(Moreno-Hernandez, Rodriguez-Reyes et al. 2010), India(D'AoÛt, Pataky et al. 2009) and Belgium(D'AoÛt, Pataky et al. 2009). Of the 15 included studies, 14 were within subject repeated measures design studies with one being a between subject comparison study (D'AoÛt, Pataky et al. 2009) which comprised 3 subject groups: habitually barefoot, habitually minimally shod and habitually shod. The ages of participants in the studies ranged from 5-74 years old; however only two (Sacco, Akashi et al. 2010, Tsai and Lin 2013) of the fifteen studies assessed participants of 50 years of age or older. Five studies (Oeffinger, Brauch et al. 1999, Wolf, Simon et al. 2008, Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Chard, Greene et al. 2013) investigated differences between footwear in children under the age of 13, with the remainder investigating the response of young-middle aged adults to barefoot walking. These data are summarised in Table 2.

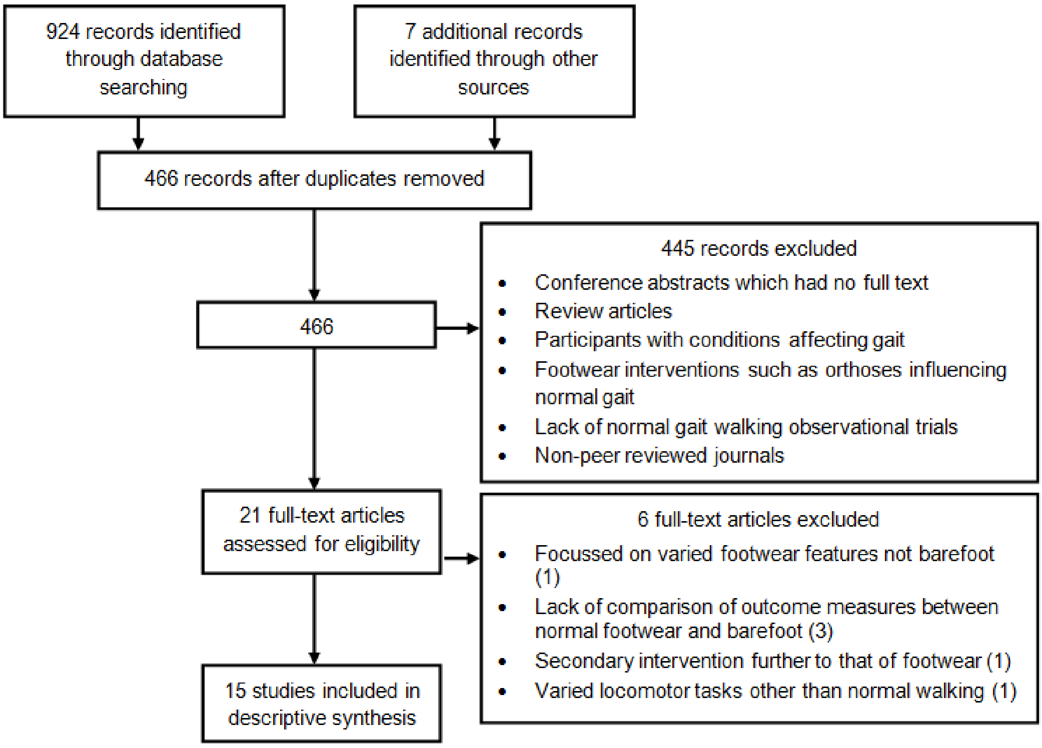

Figure 8 - Flowchart demonstrating the selection of articles through the review

Measurement Approach

Two studies set a consistent gait speed with ten studies allowing for participants to self-select their velocity. Two studies monitored gait speed and then matched their gait speed on the treadmill and one study neither reported gait velocity nor acknowledged if it was self-selected or fixed. Fourteen of the fifteen studies analysed the differences between walking barefoot and wearing various types of footwear whereas the study by D'Août et al (D'AoÛt, Pataky et al. 2009) was novel in its approach of comparing habitually barefoot walkers with two different habitually shod populations. Of the fourteen studies analysing walking barefoot in comparison to walking in footwear three studies investigated athletic shoes, two explored flip-flops and sandals and one investigated the effects of just socks. Five studies compared the effects of more than one type of footwear to barefoot with five studies being unclear about the type of shoes used in the studies. Seven of the fourteen studies used a standardised shoe across participants and in the case of the study by Wirth et al (Wirth, Hauser et al. 2011) the flexible shoe condition was standardised but the conventional shoes were not. Eight studies collected spatial-temporal data, six studies assessed kinetic variables, five studies reported kinematic data, three studies used electromyography to study muscle activity patterns and one study used ultrasound to explore muscle contractile behaviour.

Spatial-temporal Variables

Walking barefoot compared to shoes results in a reduced step and/or stride length (Oeffinger, Brauch et al. 1999, Wolf, Simon et al. 2008, Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Keenan, Franz et al. 2011, Wirth, Hauser et al. 2011). This reduction is limited when walking in more flexible footwear (Wirth, Hauser et al. 2011) and reversed in older adults when walking in socks (Tsai and Lin 2013). Walking barefoot was shown to correspond to an increase in cadence (Wolf, Simon et al. 2008, Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Wirth, Hauser et al. 2011) and similarly this difference was limited in more flexible footwear (Wirth, Hauser et al. 2011). The difference failed to reach significance in the study by Oeffinger et al (Oeffinger, Brauch et al. 1999), but an increase was observed (134 steps/min barefoot, 126 steps/min in shoes). Other variables which showed significant differences from shoes to barefoot were that of percentage double support time decreasing (Lythgo, Wilson et al. 2009), stance time decreasing (Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Zhang, Paquette et al. 2013) swing time increasing (Moreno-Hernandez, Rodriguez-Reyes et al. 2010) and stride time decreasing (Wolf, Simon et al. 2008, Lythgo, Wilson et al. 2009). Gait velocity differences between conditions was variable with some studies noting a decrease in velocity when barefoot (Lythgo, Wilson et al. 2009, Moreno-Hernandez, Rodriguez-Reyes et al. 2010, Wirth, Hauser et al. 2011) and some showing no significant differences (Oeffinger, Brauch et al. 1999, Wolf, Simon et al. 2008, Tsai and Lin 2013). Older adults (mean age 74.60 ±7.21years) were observed to reduce their gait velocity when walking in socks compared to barefoot [144]. The data is summarised in Table 3.

	Tsai & Lin (Tsai and Lin 2013)	al (Lythgo, Wilson et	al (Morio, Lake et al. 2009)	(Wolf, Simon et al. 2008)	et al (Keenan, Franz et al. 2011)	(Oeffinger, Brauch et al.	al (D'AoÛt, Pataky et	al (Chard, Greene et al. 2013)	al (Scott, Murley et al. 2012)	(Zhang, Paquette et al.	al (Sacco, Akashi et al. 2010)	Barrett (Carl and Barrett 2008)	(Cronin and Finni 2013)	al (Wirth, Hauser et al. 2011)	Hernandez, Rodriguez-Reyes
Title and Abstract	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
Introduction	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2	11= 2
Methods	10111101=6	11011001=5	10111001=5	10011001=4	10101101=5	10001101=4	10111-01=6	10111111=7	10111101=6	10101111=6	10111001=5	10001001=3	10111101=6	10111101=6	10111001=5
Results	011101=4	111001=4	011001=3	011100=3	011101=4	110000=2	110111=5	111101=5	011111=5	011101=4	111111=6	010101=3	111101=5	010101=3	111101=5
Discussion	111=3	111=3	111=3	101=2	111=3	111=3	111=3	111=3	111=3	111=3	111=3	111=3	111=3	111=3	101=2
Total Score	**16/20**	**15/20**	**14/20**	**12/20**	**15/20**	**12/20**	**16/19**	**18/20**	**17/20**	**16/20**	**17/20**	**12/20**	**17/20**	**15/20**	**15/20**

Table 1 - Results of the Methodological Quality Assessment following the critical appraisal tool developed based upon the STROBE Checklist for reporting observational studies and the CASP appraisal tool. The checklist offers a list of recommendations which should be included to ensure that all required steps for successful scientific reporting were taken and if the relevant information is presented clearly. A 1 in the table illustrates that this criteria was satisfied and a 0 demonstrates that this was missing.

	Sample Size	Age	Outcome Measures	Conditions	Test Order Randomised	Standardised shoes
Tsai & Lin (2013)	41 (21 young adults, 20 older adults)	22.52 ± 2.48 years and 74.60 ± 7.21 years	Spatial-temporal	Barefoot and socks	Yes	No
Lythgo et al (2009)	898 children, 82 young adults	5-13 years and 19.62 ± 1.60 years	Spatial-temporal	Barefoot and athletic shoes	No	No
Morio et al (2009)	10 young adults	25.4 ± 6.4 years	Kinematics – forefoot-rearfoot relative motion	Barefoot and 2 sandals (hard and soft sole)	No	Yes
Wolf et al (2009)	18 children	8.2 ± 0.7 years	Kinematics – foot motion, spatial-temporal	Barefoot, conventional and flexible shoes	No	Yes
Keenan et al (2011)	68 adults	34 ± 11 years	Kinetics, spatial-temporal	Barefoot and various athletic shoes	Yes	Yes
Oeffinger et al (1999)	14 children	7-10 years	Kinematics, spatial-temporal, kinetics	Barefoot and athletic shoes	Yes	No
D'Aout et al (2009)	255 adults (barefoot indian (BI), shod indian (SI) and shod western (W))	BI: 46.3±14.9 years, SI: 34.3±11.5 years and W: 33.9±13.1 years	Kinetics - plantar pressures	Habitual barefoot vs Habitual Shod during barefoot walking	N/A	N/A
Chard et al (2013)	13 children	10.3 ± 1.6 years	Kinematics	Barefoot and flip-flops	Yes	Yes
Scott et al (2012)	28 young adults	21.2 ± 3.8 years	Electromyography	Barefoot, athletic and flexible shoe	Yes	Yes
Zhang et al (2013)	10 young adults	25.8 ± 4.83 years	Kinematics, Kinetics, spatial-temporal	Barefoot, flip-flops, sandals and athletic shoes	Yes	Yes
Sacco et al (2010)	21 healthy adults	50.9 ± 7.3 years	Kinetics, electromyog	Barefoot and Habitual Shoes	No	No

			raphy			
Carl & Barrett (2008)	10 young adults	24.6 years	Kinetics – plantar pressures	Barefoot (socks), flip-flops and athletic shoes	No	Yes
Cronin & Finni (2013)	10 adults	29 ± 4 years	Spatial-temporal, lower limb muscle fascicle behaviour	Barefoot and shoes (unknown type)	Yes	No
Wirth et al (2011)	30 adults	31.4 ± 12.8 years	Electromyography, spatial-temporal	Barefoot, conventional shoes and flexible shoes	Yes	Conventional – No, Flexible - Yes
Moreno-Hernandez et al (2010)	120 children	6-13 years	Spatial-temporal	Barefoot and school uniform shoes	No	No

Table 2- Summary of study characteristics of articles included in review.

Kinematic Variables

There are considerable kinematic differences observed in various studies particularly with respect to changes in foot motion. Forefoot width and forefoot spreading under load during walking is significantly increased when barefoot compared to shoes (Wolf, Simon et al. 2008) and sandals (Morio, Lake et al. 2009) in populations used to walking in shoes. There are also significantly reduced medial longitudinal arch length changes in shoes compared to barefoot (Wolf, Simon et al. 2008). In addition, anatomically, habitual barefoot walkers have been shown to have considerably wider feet than their shod counterparts, and this is particularly prevalent in the forefoot region (D'AoÛt, Pataky et al. 2009). Walking barefoot also led to a change in the ankle angle at initial contact with a significant increase in plantarflexion corresponding to a flatter foot placement compared to athletic shoes, sandals and flip-flops (Oeffinger, Brauch et al. 1999, Morio, Lake et al. 2009, Chard, Greene et al. 2013, Zhang, Paquette et al. 2013). Other variables of foot motion revealing differences between footwear are reduced eversion, adduction, external rotation and foot torsion when wearing shoes or sandals (Oeffinger, Brauch et al. 1999, Wolf, Simon et al. 2008, Morio, Lake et al. 2009). Aside from differences observed purely in the foot and ankle motion, footwear also appears to alter knee kinematics. An increase in knee flexion is observed at contact when walking barefoot (Oeffinger, Brauch et al. 1999, Zhang, Paquette et al. 2013) but a greater knee and ankle ROM exists throughout stance when wearing footwear (Zhang, Paquette et al. 2013). The summary of kinematic variables is displayed in Table 4.

Kinetic variables

The kinetic variables described in the literature are quite varied and findings are at times contradictory between studies. For example, Oeffinger and colleagues (Oeffinger, Brauch et al. 1999) observed an increased hip extensor moment at terminal swing and decreased knee flexor moment at weight acceptance when walking barefoot compared to athletic shoes, whereas the opposite was observed in the study by Keenan and colleagues (Keenan, Franz et al. 2011). Keenan et al (Keenan, Franz et al. 2011) also reported a reduced hip flexor moment when walking barefoot which was supported by Zhang et al (Zhang, Paquette et al. 2013). Other variables which demonstrated significant differences between footwear conditions were a reduced initial peak vertical ground reaction force (GRF) (Sacco, Akashi et al. 2010, Keenan, Franz et al. 2011) reduced drop in force between primary and secondary vertical impact peaks (Sacco, Akashi et al. 2010), reduced braking GRF (Sacco, Akashi et al. 2010, Keenan, Franz et al. 2011) and reduced propulsive GRF (Sacco, Akashi et al. 2010) when walking barefoot. These correspond to a decreased ankle dorsiflexor moment in early stance (Zhang, Paquette et al. 2013) and reduced ankle plantar flexor moments in late stance (Oeffinger, Brauch et al. 1999) which were also reported during barefoot walking. On the other hand Keenan et al (Keenan, Franz et al. 2011) and Zhang et al (Zhang, Paquette et al. 2013) observed an increase in propulsive force when barefoot compared to athletic shoes. A decreased knee varus moment (Keenan, Franz et al. 2011) and greater ankle inversion moment at late stance (Zhang, Paquette et al. 2013) were also reported when walking barefoot. Aside from joint forces and moments, plantar pressure and Centre of Pressure (COP) displacement data was also reported. Peak plantar pressures were reported to be increased when walking barefoot compared to in athletic shoes and flip flops under the calcaneus and metatarsal heads but there was no difference observed under the hallux region [24]. Peak plantar pressures and pressure impulses were observed to be lowest in habitually barefoot walkers under the heel and metatarsal regions (D'AoÛt, Pataky et al. 2009). There were however lower relative peak plantar pressures witnessed under the toe and midfoot regions in the Western habitually shod group (D'AoÛt, Pataky et al. 2009). In terms of COP displacement, larger mediolateral but reduced anteroposterior displacements were observed when walking barefoot compared to flip-flops, sandals and athletic shoes (Zhang, Paquette et al. 2013). Data is summarised in Table 5.

	Study	Conditions	Results
Velocity Sig. slower when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Wirth et al (Wirth, Hauser et al. 2011) Moreno-Hernandez et al (Moreno-Hernandez, Rodriguez-Reyes et al. 2010)	Barefoot or Athletic Shoes Barefoot (BF), Normal Shoes (NS) Barefoot (BF), School Shoes (SS)	Mean reduction of 8cm/s barefoot BF: 0.04m/s slower than NS (p=0.001). BF: 113.32cm/s(19.52), SS:118.69cm/s(18.13) (p<0.001)
Sig. slower in socks	Tsai & Lin (Tsai and Lin 2013)	Old in socks (S) or barefoot (BF)	Old: BF: 92.51cm/s (19.18), S:80.76cm/s (23.12) * (p<0.05)

No Sig. difference between footwear	Wolf et al (Wolf, Simon et al. 2008) Oeffinger et al (Oeffinger, Brauch et al. 1999) Wirth et al (Wirth, Hauser et al. 2011) Tsai & Lin (Tsai and Lin 2013)	Barefoot (BF), normal shoes (NS) or flexible shoes (FS) Barefoot (BF), athletic shoes (AS) Barefoot (BF), Flexible Shoes (FS) Young in socks (S) or barefoot (BF)	BF:1.29m/s[0.14], NS:1.28m/s[0.13], FS:1.31m/s[0.15] (p=0.679) BF: 139.11cm/s(16.87), AS:143.42cm/s(14.61) (p=0.512) BF: 0.01m/s slower than FS (p=0.25) Young: BF: 101.32cm/s (14.26), S:101.12cm/s (13.86),
Step length Sig. shorter when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Wirth et al (Wirth, Hauser et al. 2011) Moreno-Hernandez et al (Moreno-Hernandez, Rodriguez-Reyes et al. 2010)	Barefoot or Athletic Shoes Barefoot (BF), Normal Shoes(NS), Flexible Shoes(FS) Barefoot (BF), School Shoes (SS)	Mean reduction of 5.5cm barefoot BF: 0.03 less than NS, BF:0.01 less than FS (p<0.001) (m) BF:56.35(6.74), SS:60.05(6.92) (p<0.001) (cm)
Stride length Sig. shorter when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Wolf et al (Wolf, Simon et al. 2008) Keenan et al (Keenan, Franz et al. 2011) Oeffinger et al (Oeffinger, Brauch et al. 1999)	Barefoot or Athletic Shoes Barefoot (BF), normal shoes (NS) or flexible shoes (FS) Barefoot (BF), athletic shoes (AS) Barefoot (BF), athletic shoes (AS) Young and Old, in socks (S) or barefoot (BF)	Mean reduction of 11.1cm barefoot BF:1.17m*[0.10], NS:1.24m[0.09], FS:1.23mm[0.11] (p=0.001) BF:2.15m(0.32), AS:2.29m(0.29) (p<0.001) BF:125.40cm(13.55), AS:137.18cm(11.49) (p=0.032) Young: BF: 67.63(6.31), S:66.99(5.96), Old: BF:67.80(9.30), S:62.87(12.06)* (p<0.05) (%height)

	Tsai & Lin (Tsai and Lin 2013)		
Cadence Sig. faster when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Wolf et al (Wolf, Simon et al. 2008) Wirth et al (Wirth, Hauser et al. 2011) Moreno-Hernandez et al (Moreno-Hernandez, Rodriguez-Reyes et al. 2010)	Barefoot or Athletic Shoes Barefoot (BF), normal shoes (NS) or flexible shoes (FS) Barefoot (BF), Normal Shoes(NS), Flexible Shoes (FS) Barefoot (BF), School Shoes (SS)	Mean increase of 3.9steps/min barefoot BF:132.2[8.9], NS:123.5[7.6], FS:127.6 [7.56](p=0.001) BF: 2.93steps/min more than NS, BF: 1.45steps/min more than FS (p<0.001) BF:122.48steps/min(13.83), SS:118.97steps/min(14.35) (p<0.001)
Double Support time Sig. reduced when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009)	Barefoot or Athletic Shoes	Mean reduction of 1.6% of gait cycle when barefoot
Stance time Sig. reduced when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Zhang et al (Zhang, Paquette et al. 2013) Moreno-Hernandez et al (Moreno-Hernandez, Rodriguez-Reyes et al. 2010)	Barefoot or Athletic Shoes Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS) Barefoot (BF), School Shoes (SS)	Mean reduction of 0.8% barefoot BF:0.70s*(0.02), FF:0.73s(0.02), S:0.74s(0.02), AS:0.77s# (0.03) (p=0.0001) *sig. less than FF,S and AS, #sig. more than FF and S BF:56.30%gait cycle(1.62), SS:57.04%gait cycle(3.03) (p=0.007)
Swing	Moreno-Hernandez et al (Moreno-	Barefoot (BF), School Shoes	BF:43.71%gait cycle(1.62), SS:42.97%gait cycle(3.04)

time Sig. increased when barefoot	Hernandez, Rodriguez-Reyes et al. 2010)	(SS)	(p=0.006)
Stride time Sig. reduced when barefoot	Lythgo et al (Lythgo, Wilson et al. 2009) Wolf et al (Wolf, Simon et al. 2008)	Barefoot or Athletic Shoes Barefoot (BF), normal shoes (NS) or flexible shoes (FS)	Mean reduction of 25ms barefoot BF:0.91s[0.06], NS:0.98s[0.06], FS:0.94s[0.06] (p<0.001)

Table 3 – Summary of the spatial-temporal variables. Results are displayed as group means followed by standard deviations in parentheses () or standard error measurement in brackets [].

	Study	Conditions	Results
Forefoot Width/ Spreading Sig. increase in forefoot spreading when walking barefoot Habitual barefoot walkers have sig. wider feet	Wolf et al (Wolf, Simon et al. 2008) Morio et al (Morio, Lake et al. 2009) D'Aout et al (D'AoÛt, Pataky et al. 2009)	Barefoot (BF), normal shoes (NS) or flexible shoes (FS) Barefoot (BF), Hard Sandal (HS), Soft Sandal (SS) Habitually barefoot Indian (BI) vs habitually shod Indian (SI) vs Western Shod (WS)	BF:9.7%(3.1), NS:4.3%(1.4), FS:5.9%(1.4) (p<0.001) (% change from standing) Metatarsal heads BF: 1.1°(0.8), SS: 0.9°(1.3), HS: 1.6°(1.7) (change from standing calibration SS: -2.5°(1.8), HS: -3.5°(1.8)) Metatarsal bases BF: 0.8°(1.0), SS: 1.1°(1.1), HS: 1.7°(1.4) (change from standing calibration SS: -0.7°(0.9), HS: -2.4°(1.4)) BI: approx. 37%, SI: approx. 35.5%, WS: approx. 33.5% (p=0.000) (%footwidth/length) Foot area normalized to stature squared = WS: 15.5% smaller than both Indian groups (p=0.000)
Ankle Angle at initial	Morio et al (Morio, Lake et al. 2009)	Barefoot (BF), Hard Sandal (HS), Soft Sandal (SS)	BF: 4.8°(2.1), SS: 5.5°(1.5), HS: 6.8°(2.3) (p<0.05) (dorsiflexion excursion)

contact (sagittal plane) Sig. more plantarflexed when barefoot	Oeffinger et al (Oeffinger, Brauch et al. 1999) Chard et al (Chard, Greene et al. 2013) Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF), athletic shoes (AS) Barefoot (BF) and flip-flops (FF) Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	A decrease of 3° in ankle plantarflexion in AS ($p<0.05$) BF: 1.1°(8.3) FF: 12.0°(12.2) ($p=0.005$) BF:-3.9°(3.9) FF:0.4°(5.0) S:-0.1°(4.5) AS:3.7°(3.8) ($p=0.001$)
Foot Angle at Contact (sagittal plane) Sig. more plantarflexed when barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:19.2°(3.4) FF:25.5°(3.9) S:24.9°(3.6) AS:29.5°(4.5) ($p<0.001$) (angle between foot and ground)
Foot Eversion Sig. more when barefoot	Morio et al (Morio, Lake et al. 2009)	Barefoot (BF), Hard Sandal (HS), Soft Sandal (SS)	BF: 9.5°(2.9), SS: 8.2°(2.8), HS: 7.9°(2.7) ($p<0.05$)
Foot Adduction Sig. more when barefoot	Morio et al (Morio, Lake et al. 2009)	Barefoot (BF), Hard Sandal (HS), Soft Sandal (SS)	BF: 11.5°(1.8), SS: 9.8°(2.0), HS:8.3°(1.6) ($p<0.05$)
Foot External rotation Sig. more when barefoot	Wolf et al (Wolf, Simon et al. 2008)	Barefoot (BF), normal shoes (NS) or flexible shoes (FS)	BF:20.9%[3.9], NS:18.7%[4.3], FS:19.4%[4.7] ($p<0.001$) (% change from standing)
Foot torsion Sig. more	Wolf et al (Wolf, Simon et al. 2008)	Barefoot (BF), normal shoes (NS) or flexible shoes (FS)	BF: 9.8°[3.0], NS: 4.7°[1.6], FS: 5.2°[2.0] ($p<0.001$) (forefoot-hindfoot relative motion in

when barefoot			transverse plane)
Medial Longitudinal Arch Sig. greater change in length when barefoot	Wolf et al (Wolf, Simon et al. 2008)	Barefoot (BF), normal shoes (NS) or flexible shoes (FS)	BF:9.9%[2.5], NS:5.9%[1.5], FS:6.0%[1.8] (p<0.001) (% change from standing)
Knee Flexion at initial contact Sig. more flexion when barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:-8.0°(3.9) FF:-6.3°(3.7) S:-6.3°(3.9) AS:-5.2°(3.4) (p=0.001) (negative value means greater flexion)
Knee ROM throughout stance Sig. reduced when barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF: 39.9°(5.3) FF:44.2°(4.7) S:45.8°(4.8) AS:46.7°(4.4) (p<0.001)
Ankle plantarflexion ROM in late stance Sig. reduced when barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF: 8.0°(1.9) FF:8.7°(1.4) S:9.4°(1.7) AS:11.8°(2.9) (p=0.001)

Table 4– Summary of the kinematic variables. Results are displayed as group means followed by standard deviations in parentheses () or standard error measurement in brackets [].

Muscle activity and behaviour

Three studies used electromyography measures to determine differences in muscle activity patterns during walking in footwear and barefoot. The variables of interest included mean peak amplitude (the maximum amplitude within a stride and averaged across 8 ipsilateral steps, reported in mV (Scott, Murley et al. 2012) or % of MVC (Wirth, Hauser et al. 2011)), time to peak amplitude (reported in terms of % of gait cycle (Sacco, Akashi et al. 2010, Scott, Murley et al. 2012)) and maximum peak amplitude (reported as % of MVC (Wirth, Hauser et al. 2011)). Scott et al (Scott, Murley et al. 2012) stated that tibialis anterior (TA) displayed a significantly reduced mean peak amplitude from shoes (0.12mV) to barefoot (0.09mV) ($p<0.001$), an increase in peroneus longus (PL) mean peak amplitude (0.17mV barefoot, 0.13mV flexible shoe, 0.14mV stability shoe) ($p<0.05$) and no difference in the medial gastrocnemius (MG) (0.06mV for all three footwear). However the time to peak amplitude occurred later in the TA (6.02% to 5.53%) ($p=0.008$) and PL (50.11% to 47.55%) ($p=0.004$) barefoot compared to the flexible shoe and stability shoe respectively and occurred earlier in the MG compared to the stability shoe (41.58% to 43.80%) ($p<0.001$) (Scott, Murley et al. 2012). Conversely, Sacco et al (Sacco, Akashi et al. 2010) demonstrated that although not significant ($p=0.06$) there was a trend toward the peak amplitude in the TA occurring later in shoes (CG: 5.46% to 6.52% DG: 5.61% to 6.58%). Sacco et al (Sacco, Akashi et al. 2010) also reported that the Vastus Lateralis time to peak amplitude occurred significantly ($p=0.002$) earlier when barefoot (CG: 10.76% to 15.47% DG: 14.14% to 15.35%). It is worth noting however that these statistics comprise both the control and diabetic group data and there is no statistical test reported stating whether these groups are similar. A slightly higher mean amplitude was observed in various back muscles (Lumbar Iliocostalis $p=0.015$, Sternocleidomastoideus $p=0.008$) and neck extensor muscles ($p=0.003$) when barefoot compared to conventional shoes (Wirth, Hauser et al. 2011). The same tendency was observed in the Lumbar Longissimus, the Lumbar Multifidi and Trapezius Pars Descendens; however these did not reach statistical significance. In comparison to the flexible shoe condition mean amplitude was only significantly higher barefoot in the Sternocleidiomastoideus (p=0.01) (Wirth, Hauser et al. 2011). Wirth et al (Wirth, Hauser et al. 2011) also reported that the maximum amplitude in the Neck Extensor muscles exhibited a significantly higher (p=0.02) amplitude barefoot than in conventional shoes. These differences although significant are relatively marginal in absolute terms with the mean activity ranging from a change of 0.23-0.47% of the maximum voluntary contraction. With no effect sizes being reported it is difficult to comment on the strength of the difference. Cronin & Finni (Cronin and Finni 2013) found no significant differences in soleus or medial gastrocnemius fascicle length or velocity changes between footwear despite seeing significant differences in the spatial-temporal characteristics of gait.

	Study	Conditions	Results
Hip Extensor Moment Sig. reduced when barefoot	Keenan et al (Keenan, Franz et al. 2011)	Barefoot (BF), 2x athletic shoes (AS)	BF: 0.48Nm/Kgm(0.13) AS:0.51Nm/Kgm(0.14) (p<0.003)
Hip Flexor Moment Sig. reduced when barefoot	Keenan et al (Keenan, Franz et al. 2011) Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF), 2x athletic shoes (AS) Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:0.35Nm/Kgm (0.14) AS:0.50Nm/Kgm(0.15) (p<0.003) BF:0.63(0.09), FF:0.66(0.10), S:0.67(0.11), AS:0.66(0.11) (p=0.007) (Nm/Kg)
Knee Flexor Moment Sig.	Keenan et al (Keenan, Franz et al. 2011)	Barefoot (BF), 2x athletic shoes (AS)	BF:0.11Nm/Kgm(0.09) AS1:0.07Nm/Kgm(0.09) AS2:0.05Nm/Kgm (0.08)

increased when barefoot			($p<0.003$)
Ankle Dorsiflexor moment - early stance Sig. reduced when barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:0.11(0.04), FF:0.11(0.04), S:0.13(0.04), AS:0.16(0.04) ($p=0.008$) (Nm/Kg)
Initial Peak vGRF Sig. reduced when barefoot	Keenan et al (Keenan, Franz et al. 2011) Sacco et al (Sacco, Akashi et al. 2010)	Barefoot (BF), 2x athletic shoes (AS) Barefoot (BF) and Shoes (SH)	BF:109.94%BW(7.53) AS:112.37%BW(7.26) ($p<0.003$) BF:1.04(0.09) SH:1.09(0.09) ($p<0.001$) (times BW)
Braking GRF Sig. reduced when barefoot	Keenan et al (Keenan, Franz et al. 2011) Sacco et al (Sacco, Akashi et al. 2010)	Barefoot (BF), 2x athletic shoes (AS) Barefoot (BF) and Shoes (SH)	BF: 17.59%BW(3.84) AS1:18.73%BW(4.08) AS2:18.80%BW(3.99) ($p<0.003$) BF: -0.131(0.02) SH: -0.142 (0.04) ($p<0.001$) (times BW)
Propulsive GRF Sig. increased barefoot	Keenan et al (Keenan, Franz et al. 2011)	Barefoot (BF), 2x athletic shoes (AS)	BF: 20.09%BW(3.43) AS1:18.42%BW (3.22) AS2:19.17%BW(3.05) ($p<0.003$)
Sig. decreased barefoot	Sacco et al (Sacco, Akashi et al. 2010)	Barefoot (BF) and Shoes (SH)	BF: 0.155(0.02) SH: 0.178 (0.02) ($p<0.001$) (times BW)
Knee Varus Moment Sig. decreased barefoot	Keenan et al (Keenan, Franz et al. 2011)	Barefoot (BF), 2x athletic shoes (AS)	BF: 0.31Nm/Kgm(0.06) AS:0.34Nm/Kgm(0.07) ($p<0.003$)

Ankle Inversion Moment – late stance Sig. increased barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:0.29(0.23), FF:0.26(0.22), S:0.26(0.22), AS:0.17(0.10) (p=0.026) (Nm/Kg)
Peak Plantar Pressure Sig. greater barefoot (in habitually shod subjects)	Carl & Barrett (Carl and Barrett 2008)	Barefoot, Flip-flops, athletic shoes	Barefoot > Flip-flops > Shoes (no values) under metatarsals and calcaneus
Sig. reduced in habitual barefoot walkers compared to habitual shod walkers	D'Aout et al (D'AoÛt, Pataky et al. 2009)	Habitually Barefoot Indian (BI) vs Shod Indian (SI) vs Western Shod (WS)	BI < SI < WS under heel and metatarsals (no values)
COP displacement (Medio-lateral) (cm) Sig. greater barefoot	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:5.5(1.4), FF:4.7(1.2), S:4.5(1.1), AS:4.0(1.0) (p=0.009)
COP displacement (Antero-posterior) (cm) Sig. reduced	Zhang et al (Zhang, Paquette et al. 2013)	Barefoot (BF),flip-flops (FF), sandals (S) and athletic shoes (AS)	BF:21.1(1.3), FF:26.2(2.1), S:26.8(1.6), AS:26.8(2.2) (p=0.0001)

barefoot			

Table 5– Summary of the kinetic variables. Results are displayed as group means followed by standard deviations in parentheses () or standard error measurement in brackets [].

Discussion

The aim of this systematic review was to explore the research on walking barefoot to help understand the effect that wearing footwear has on gait kinematics, kinetics and muscular activity. It was also possible to compare different footwear types and how closely they link to walking barefoot in terms of the variables of interest.

A marked discrepancy in the results of some studies is observed with respect to changes in gait velocity between footwear conditions, with some studies reporting a reduction in velocity when barefoot and some reporting no significant difference. This is potentially explained by the familiarity or variability of the footwear used. A standardised shoe was employed in the studies finding no significant differences, whereas in the studies that noted a decrease in gait velocity when barefoot participants used their own habitual shoes. This suggests that the familiarity of the shoes worn by participants potentially has a significant impact on gait parameters such as gait velocity and thus future studies investigating such parameters should take this into consideration when designing their methodology.

Results were more conclusive regarding step and/or stride length differences with many studies observing a clear reduction when walking barefoot. Some authors suggest this could be due to a pendulum lengthening effect such that the extra weight of the shoe leads to greater inertial load during the swing phase and a corresponding increase in step length (Oeffinger, Brauch et al. 1999). This reduction in stride length was shown to be limited in more flexible shoes. The weights of these were not reported. However, based on the descriptions of the footwear and the type of footwear they were compared against, it can be assumed that these were significantly lighter, supporting the pendulum-lengthening suggestion. On the other hand, the change in stride length may not be purely due to increased weight as Tsai & Lin (Tsai and Lin 2013) observed a reduction in the stride length from barefoot to socks in older adults. This suggests that in this population a mechanism other than that of distal weight influenced their gait performance to bring about a change in their stride length. Interestingly, no difference was observed in younger adults suggesting that wearing only socks influences those with reduced gait performance to a greater extent. It could be proposed therefore that the observed changes in stride length are as a result of a change in gait concerned with gait stability as opposed to purely an inertial difference.

A significant increase in ankle plantarflexion was observed when walking barefoot in a number of studies resulting in a flatter foot placement at contact (Oeffinger, Brauch et al. 1999, Morio, Lake et al. 2009, Chard, Greene et al. 2013, Zhang, Paquette et al. 2013) which also corresponds to a delayed and reduced mean peak amplitude of the Tibialis Anterior (TA) (Scott, Murley et al. 2012). This increase in plantarflexion when barefoot could be explained as a method of increasing the surface area of the foot at contact. Following the formula P = F/A, where P = pressure, F = force and A = contact surface area; an increase in the area of the foot at contact would serve to reduce the pressure, and potential associated discomfort, experienced at that point. This was demonstrated in the study by D'Août and colleagues (D'AoÛt, Pataky et al. 2009). The authors compared habitual barefoot walkers, who have never worn shoes, with habitually shod subjects, who wear shoes on a daily basis outdoors, and also incorporated an intermediate group of habitually minimally shod users who normally wear open-toed footwear such as flip-flops or sandals but walked mostly barefoot as a child in accordance with native habits. They used plantar pressure plates to analyse the long term effects that footwear use has on foot function and foot shape during repeated barefoot walking trials. They observed that the habitual barefoot walkers, displaying an anatomically larger plantar foot area, had significantly reduced peak plantar pressures at the heel and metatarsal regions compared to the habitually shod populations. This suggests that due to a larger plantar surface area the habitually barefoot walkers are able to distribute the pressures more evenly across the foot. Additional to the anatomical differences in foot area, the habitual barefoot walkers were also observed to adopt a flatter initial foot placement thus further allowing for distribution of pressures across a larger area. D'Août et al also state that a flatter foot placement allows for the pressures to be distributed over a longer period of time, reducing the pressure impulse, instead of being applied quickly at one point at initial contact and then relatively low pressures following this as observed in shod populations (D'AoÛt, Pataky et al. 2009). Clearly these findings from the participants in the study by D'Aout et al suggest that this is as a result of habitual lack of footwear (D'AoÛt, Pataky et al. 2009). However, it must be noted that there are other differences besides the lack or presence of footwear between these populations. The habitually shod population have grown up in a Westernised environment compared to the native rural environment of the habitual barefoot walkers. They are therefore clearly not of the same population and thus comparisons drawn between them must be taken with caution. Other factors such as walking surface (roughness and compliance), stature or age, which were different between the groups, could also account for a portion of the variance observed other than that of simply the lack of footwear worn. However the authors are aware of this issue and discuss it in their paper whilst highlighting the need for similar population groups as a suggestion for future work. Nonetheless, it is interesting to see a similar change in foot kinematics being observed in shod populations following a brief switch to walking barefooted thus suggesting it may be an inherent response. Even so, it must be stated that for those who are accustomed to wearing shoes, walking barefoot results in increased plantar pressures at the heel and metatarsal regions compared to walking in shoes or flip-flops (Carl and Barrett 2008).

Of note however is the observed reduction in the initial vertical peak ground reaction force witnessed in habitually shod participants walking barefoot (Keenan, Franz et al. 2011),(Sacco, Akashi et al. 2010). Furthermore there was a reduced drop between vertical GRF peaks, suggesting that forces were distributed throughout the stance period more when barefoot as opposed to a greater initial impulse followed by a reduction in force prior to a secondary steep rise to the second peak GRF. This is similar to that suggested previously from the plantar pressure data in the habitually barefoot participants (D'AoÛt, Pataky et al. 2009). This indicates that acute and long term exposure to barefoot walking changes the kinematics and associated kinetics such that forces are spread more evenly over time. This is also consistent with barefoot running literature whereby a smooth force profile is observed when running barefoot. An abundance of research (De Wit, De Clercq et al. 2000, Divert, Mornieux et al. 2005, Squadrone and Gallozzi 2009, Lieberman, Venkadesan et al. 2010) has observed that in barefoot runners there is the absence of, or a distinct reduction in, the impact force transient contrasting that of when running in shoes. It is explained by a more plantarflexed foot strike when running barefoot which enables better use of the Windlass mechanism (which is the ability of the plantar fascia, running from the base of the calcaneus to the phalanges, to raise the medial longitudinal arch by creating tension throughout stance, preventing arch collapse, and assisting at push off through elastic energy release (Bolgla and Malone 2004, Caravaggi, Pataky et al. 2009)) and absorption of load by the lower limb musculature.

This reduction in force peaks and with the force being spread more evenly over time also appears to have an impact on the joint moments experienced by participants with differences being reported between footwear. A reduction in the hip extensor, hip flexor and knee varus moments in the early stance phase were observed when walking barefoot (Keenan, Franz et al. 2011),(Zhang, Paquette et al. 2013) which was contrasted by an increase in the knee flexor moment (Keenan, Franz et al. 2011). The authors suggest that these differences in joint moments are likely to be the result of kinematic alterations aimed at impact force attenuation such as reducing stride length (Keenan, Franz et al. 2011). These kinematic and kinetic changes may also be relevant for the progression of osteoarthritis (OA). Increased knee varus, external adduction moments and adduction moment impulse increase the load through the medial compartment of the knee which is thought to lead to the progression of OA (Foroughi, Smith et al. 2009). The knee adduction moment is defined by the GRF and the moment arm of the GRF about the knee joint centre (Reeves and Bowling 2011). Therefore, barefoot walking could be potentially significant for reducing the progression of OA by reducing the magnitude of the GRF and accompanied knee adduction moment as has been witnessed in previous study (Shakoor and Block 2006, Shakoor, Sengupta et al. 2010). It has also been explained that aside from the biomechanical changes to gait the structural raised heel and medial arch support present within the shoes also contribute to increased knee varus moments (Kerrigan, Johansson et al. 2005). This could have relevance in the recommendations of athletic shoes with a raised heel and arch support to individuals particularly at a greater risk of developing osteoarthritis.

It has been noted previously that gait varies with age, thus caution must be taken when generalising the results across age groups without primary research being undertaken in older populations. It is apparent that some of the findings from this review, particularly surrounding those aspects concerned with the progression of deleterious joint conditions as well as balance and fall risk, are most relevant to the older population. However a notable finding from the review was that very little research has been completed in older age populations. Of the 15 articles in the review only 2 included participants over the age of 50 with the majority focussed on young or middle aged adults. Thus it seems necessary for research to be replicated in this population and assess if the same responses to barefoot walking are experienced.

Research into footwear use in children is also of great interest as footwear can have a lasting impact on the developing foot. With this in mind a number of articles reported data suggesting that footwear could potentially be restricting the natural motion of the foot thus affecting its development. Morio et al (Morio, Lake et al. 2009) and Wolf et al (Wolf, Simon et al. 2008) both reported a significant increase in the forefoot width and forefoot spreading under load barefoot compared to walking in shoes(Wolf, Simon et al. 2008) and sandals (Morio, Lake et al. 2009). This demonstrates that the shoes are somewhat limiting the motion of the foot in the forefoot region and not allowing it to spread under the load and utilise its structure. In addition to potentially affecting the load bearing mechanisms of the foot, long term use of these footwear could be affecting the anatomical structure of the foot and observing the feet in older adults supports this. Research by Chaiwanichsiri et al (Chaiwanichsiri, Janchai et al. 2009) indicates the prevalence of foot problems in older adults. They report that 87% of cases suffer from at least one form of foot deformity with 45.5% exhibiting hallux valgus and in these subjects 10% of men and 20% of women also had overriding toes and 87% had callus formations as a result. This prevalence of foot problems is supported by Menz et al (Menz and Lord 2001) who also stated that 87% of cases registered at least one foot problem and indicate that foot problems are significantly associated with reduced gait performance and their risk of falling. Insufficient room in the forefoot region of shoes could somewhat explain the occurrence of these conditions, and the data from the habitual barefoot walkers (D’AoÛt, Pataky et al. 2009) exhibiting significantly wider feet and forefoot spreading also supports this. Thus, ensuring that footwear does not impact negatively upon foot development is of vital importance to reduce the prevalence of foot problems and to allow the foot to function as it would naturally.

Another difference of note between walking barefoot and in shoes was that of changes in terms of medial longitudinal arch function. Wolf et al (Wolf, Simon et al. 2008) indicated that significantly reduced changes in length were observed when walking in shoes compared to barefoot thus suggesting that shoes inhibit the windlass mechanism. This is explained by that under load at contact the arch is under pressure to flatten thus tension is created along the plantar fascia to maintain the structure. This tension then recoils at push off causing the arch to rise; reducing the distance between heel and metatarsals. The reduction in the length changes of the arch in footwear indicates that this mechanism is somewhat inhibited. Longitudinal arch differences were also observed between habitual barefoot and habitual shod populations (D'AoÛt, Pataky et al. 2009). The habitually Western shod population generally had higher arches but significantly greater variability from very low to very high arches whereas the barefoot and minimally shod groups had lower arches but very little variation. This suggests that varied footwear use causes changes within the foot's structure and can lead to the extremes in arch height which are commonly associated with foot problems. Excessively high arches reduce the area of support and Chaiwanichsiri et al (Chaiwanichsiri, Janchai et al. 2009) noted that patients with pes planus, denoted as the lack of a medial longitudinal arch, had a reduced risk of falling. The authors suggest that the greater area of support could be the reason for this. Clearly this is an extreme condition of a lower arch and comes with associated problems, such as the lack of an effective windlass mechanism and over pronation. However as highlighted in the habitual barefoot population having lower arches than their shod counterparts (D'AoÛt, Pataky et al. 2009), aspects of modern footwear design such as arch supports could be forcing our feet into unnatural positions not allowing for normal foot function and resulting in weakness. This could explain the higher variability and prevalence of arch related foot problems in the shod populations and further research into ensuring footwear is designed such that it doesn't affect foot development and function is necessary.

It must be stated however that removing shoes and walking outside purely barefoot is likely not feasible to most populations. Shoes do offer a protective surface against the likelihood of cuts, abrasions and infections from mechanical insult and debris (Menant, Steele et al. 2008, Squadrone and Gallozzi 2009). As research has suggested that flexible, lighter, minimalist footwear is more similar than normal footwear in kinematics and kinetics to that of when barefoot running (Squadrone and Gallozzi 2009) and walking (Wolf, Simon et al. 2008, Wirth, Hauser et al. 2011), footwear design should focus on finding the balance between ensuring that the foot will be protected by the shoe whilst allowing for natural foot motion and structure to be maintained.

There are certain limitations of this review namely that of a wide range of variables being reported within the studies thus the subsequent lack of the ability to complete a quantitative meta-analysis. Additionally by limiting the sample population to that of healthy with no gait impairments we were unable to observe any effects of barefoot walking in patients with disorders which could have disrupted their gait and thus our findings cannot be applied to these patient groups.

Conclusions

We have systematically reviewed studies investigating differences in gait variables between walking barefoot and in shoes and highlighted how habitually shod populations react acutely to barefoot walking and how habitual barefoot walkers vary to those who wear shoes on a daily basis. Long term use of footwear has been shown to result in anatomical and functional changes including reduced foot width and forefoot spreading under load probably due to the constraints of the shoe structure. Walking in footwear is associated with an increase in stride length and greater dorsiflexion at foot-ground contact. Lighter and more flexible footwear appears to elicit reduced differences in gait kinematics to walking barefoot. A reduced initial vertical impact force and more even distribution of pressure across the foot is experienced when walking barefoot which is likely to be as a result of a larger contact surface area achieved via a flatter foot placement. Little research on barefoot walking has been completed in adults approaching older age where foot problems and gait deficiencies are most prevalent and thus investigation into this population is required to determine the impact of barefoot walking across the lifespan.

Chapter 3

Is walking in minimalist footwear the same as walking barefoot: A kinematic and kinetic comparison across age

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Abstract

Walking barefoot (BF) yields different kinematics and kinetics when compared to walking in conventional footwear. Many shoe companies offer barefoot-like footwear (BLF) allegedly simulating being BF whilst still offering a protective surface. The aims of this study are to investigate if wearing BLF is equivalent to walking BF and if the kinematic and kinetic changes associated with walking BF or in BLF compared to conventional footwear are witnessed across age including older people who have experienced many years of conventional footwear use. Seventy healthy adults (age range 20-87) volunteered for this study. All participants walked along a 7m lane five times in four different footwear conditions (barefoot (BF), barefoot-like footwear (BLF), their own shoes (OS) and control shoes (CS)). Kinematics and kinetics were recorded in synchrony. Walking BF lead to reduced step and stride lengths, increased ankle plantar flexion at foot-ground contact and a reduced peak loading ground reaction force (GRF). These observed differences in kinematics were reduced when wearing BLF. The differences reported when walking BF or in BLF were consistent across age. Wearing BLF does not replicate walking BF potentially due to cutaneous afferent interference and the perception of protection when wearing footwear. Wearing BLF may result in a more stable gait with step lengths being reduced whilst preserving gait speed. Footwear evokes changes further up the kinematic chain and not just the foot segment. Fit and healthy individuals may still benefit from walking BF (or in BLF) regardless of the number of years spent wearing conventional footwear.

Introduction

It has been suggested that wearing shoes can have a significant impact upon foot shape and function. Particularly for the older adult population the shoes typically worn are narrower than the natural foot width, which is known to contribute to foot irregularities such as hallux valgus (Chantelau and Gede 2002),(Al-Abdulwahab and Al-Dosry 2000). Notably, the occurrence of this condition in non-shoe wearing populations is markedly reduced (Dave, Mason et al. 2015). Similarly, longitudinal arch height is more variable in a middle aged shod population compared with a habitual barefoot population suggesting footwear may interfere with the intrinsic properties of the foot (D'AoÛt, Pataky et al. 2009).

Conversely, walking barefoot (BF) is suggested to improve proprioception, heighten the sensory mechanisms and potentially increase foot and lower leg muscle strength (Lieberman 2012). Research has shown that postural sway (Priplata, Niemi et al. 2003) and gait variability (Stephen, Wilcox et al. 2012), can be reduced by increasing the sensory information to the foot sole often via insoles aimed at artificially increasing afferent input. However, there is suggestion that simply being BF and removing the shoe structure between the foot and the ground, improves the sensorimotor system (Robbins, Waked et al. 1995). Robbins et al. (1995) observed participants were better at detecting the degree of a surface slope when BF compared to athletic footwear. Furthermore, a study investigating older adults' postural balance found a significant increase in anterior-posterior sway with cushioned walking shoes compared to BF (Brenton-Rule, Bassett et al. 2011). It was suggested this increase in sway was an automatic response to the insulating effect of the footwear on the afferent receptors. As the ageing process already causes a decline in the sensitivity of the sensory mechanisms, further impairing this system by wearing footwear which limits the information available seems illogical.

Barefoot-like footwear (BLF) are shoes which consist of a flat, flexible sole with no support or cushioning and are foot-shaped allowing the foot to spread and function as it would unshod while protecting the foot from abrasions and contamination. A recent systematic review investigating the effect of footwear on walking highlighted that when BF people take shorter steps, contact the ground with a flatter foot, have a more flexed knee and also experience a lower ground reaction force (GRF) during loading (Franklin, Grey et al. 2015). It was proposed that wearing BLF limits these differences between shod and unshod walking however there were only 3 studies included which investigated a minimal shoe. It is therefore important to ascertain if BLF is a viable alternative to BF.

The aims of this exploratory study were; 1) to investigate if walking BF or in BLF share the same kinematic and kinetic changes compared to conventional footwear; 2) to investigate if the kinematics and kinetics witnessed when walking BF or in BLF are consistent across age. We hypothesised that both walking BF and in BLF would result in similar kinematic and kinetic changes compared to conventional shoe. Likewise we expected these changes when walking BF or in BLF would be consistent across age.

Methods

70 healthy adults (43 females) volunteered. Participants' ages ranged from 20-87 years old. All participants had no gait abnormalities and were able to ambulate independently. All procedures were followed as approved by the University ethics committee. The participants' demographics are displayed in Table 6.

Participants visited the laboratory with the shoes they usually select for their normal activities. Their height and weight was measured without their shoes prior to completing a timed up and go test (TUG) (Mathias, Nayal et al. 1986). The test was repeated twice BF and twice in their own shoes (OS) in a counterbalanced order and the best time from each recorded. This was completed to illustrate that the older adults recruited for the study were fit and healthy demonstrating comparable values to previously published reference values for healthy older adults (Bohannon 2006).

Retro-reflective markers were placed bilaterally at the greater trochanter, lateral epicondyle, lateral malleolus, the base of the calcaneus and the first metatarsophalangeal joint (1MPJ). When wearing footwear, markers were attached to the shoes in the same positions as determined by palpation. The foot was defined as the vector between the calcaneus and the 1MPJ, the shank as the vector between the lateral malleolus and the lateral epicondyle and the thigh as the vector between the lateral epicondyle and the greater trochanter. A 13-camera Vicon MX system (Vicon, Oxford, UK) sampling at 250Hz was used with a residual error of less than 1mm. An embedded force platform (Type 9281C, Kistler, Winterhur, Switzerland) sampling at 1000Hz situated in the centre of a 7m walking lane recorded the ground reaction forces (GRF).

Participants walked at a self-selected speed from a mark based on 3 practice trials which served to ensure 3 steps were taken prior to the lane and their right foot made contact within the force plate. To promote spontaneity participants were not instructed to contact the force plate however when unsuccessful the trial was discarded and repeated. Participants completed between 5-8 trials in each of the four randomly assigned footwear conditions. These were barefoot (BF), a barefoot-like footwear (BLF) (Product ID: 2169, Two Barefeet Boarding Co.), a control shoe (CS) (Style Code: 10001, Hobos Womens, Style Code: 50109, Hobos Mens) (Figure 13) and the participants own shoes (OS). The ground was hard with little compliance and covered with a non-slip rubber with studded dot pattern (raised by ~1mm). Kinematics and kinetics were recorded in synchrony throughout each trial.

Post-processing of the data was completed using custom-written scripts in Matlab (MATLAB, The MathWorks, Natick, MA, USA). Kinematic data was low pass filtered using a fourth-order

Butterworth filter with a cut-off frequency of 12Hz. As this study included both males and females, gait speed was normalised to height to negate the effects of pure height differences between genders influencing the results as previously observed in a similar study (Hollman, McDade et al. 2011). Stride length was determined by computing the distance in the direction of travel between two consecutive right heel ground contacts using the right heel marker position and the same procedure completed for step length between consecutive right and left heel ground contacts. Knee angle was calculated using the angle formed by the greater trochanter, lateral epicondyle of the femur and lateral malleolus and 0° being when in the anatomical standing position in full knee extension. Ankle angle was determined in two ways; 1) as the angle formed between the lateral epicondyle of the femur, the lateral malleolus and the 1MPJ and 2) the angle formed between a vertical vector from the lateral malleolus (x,y co-ordinate from lateral malleolus and z co-ordinate from lateral epicondyle of femur), the lateral malleolus and the 1MPJ. This provided both a relative and absolute measure to account for the functional influence of knee angle differences. Participants were grouped into Young <40 years (n=20), Middle >40 years and <70 years (n=30) and Old >70 years (n=20).

Multiple mixed design repeated measures Analysis of Variance (ANOVA) were used to determine differences between footwear and age. Mauchly's test of sphericity was completed and if this test was violated a Greenhouse-Geisser correction applied. Statistical analyses were performed using SPSS V.22 (IBM Corporation, Somers, NY) with significance levels at $p < 0.05$.

Results

Kinematics

Gait Speed

Gait speed, normalised to body height, showed no main effect of footwear or age but a statistically significant footwear x age interaction ($F(5.304,177.682)= 4.769$; $p <0.001$; $\eta^2_{partial}$ =

0.125). The young age group walked slower BF than in their OS (3.61% slower) but showed no differences to BLF and the CS. Conversely, in the middle and old age groups, participants walked slower BF compared with all other footwear (Middle: 3.68% slower than BLF, 4.69% slower than CS, 5.37% slower than OS; Old: 5.72% slower than BLF, 8.02% slower than CS, 7.59% slower than OS) (Figure 9).

Age Group	No. In group	Age (years)	BMI (kg/m^2)
Young	20	27.85 (4.83)	23.25 (3.46)
Middle	30	54.85 (9.85)	25.04 (3.48)
Old	20	77.55 (4.39)	25.21 (4.39)

Age Group	Mean age (years)	BMI (kg/m^2)	TUG Time Barefoot (secs)	TUG Time Shoes (secs)	Falls in last 12 months	Own shoe weight (g)
20 years	23.8 (2.78)	22.52 (3.37)	8.60 (1.21)	8.34 (0.95)	0 (0)	309.5 (53.72)
30 years	31.9 (2.23)	23.74 (3.61)	9.24 (0.97)	8.81 (0.64)	0 (0)	276.5 (86.86)
40 years	42.9 (2.30)	25.65 (4.11)	8.59 (1.82)	8.09 (1.38)	0.3 (0.67)	348.5 (91.68)
50 years	53.7 (2.94)	24.72 (3.29)	8.55 (1.27)	8.77 (1.28)	0 (0)	318 (106.07)
60 years	65.5 (3.27)	24.90 (3.31)	9.99 (1.20)	9.44 (0.87)	0.3 (0.48)	290.5 (145.38)
70 years	74 (2.75)	25.01 (3.16)	9.34 (1.69)	8.76 (1.50)	0.9 (1.20)	300 (72.84)
80 years	81.1 (2.28)	25.56 (2.60)	10.92 (1.71)	10.37 (1.82)	0.7 (1.25)	308.5 (70.36)

Table 6 - A summary of the participant's statistics. Data is displayed as means (s.d). BMI = Body Mass Index, TUG = Timed Up and Go Test.

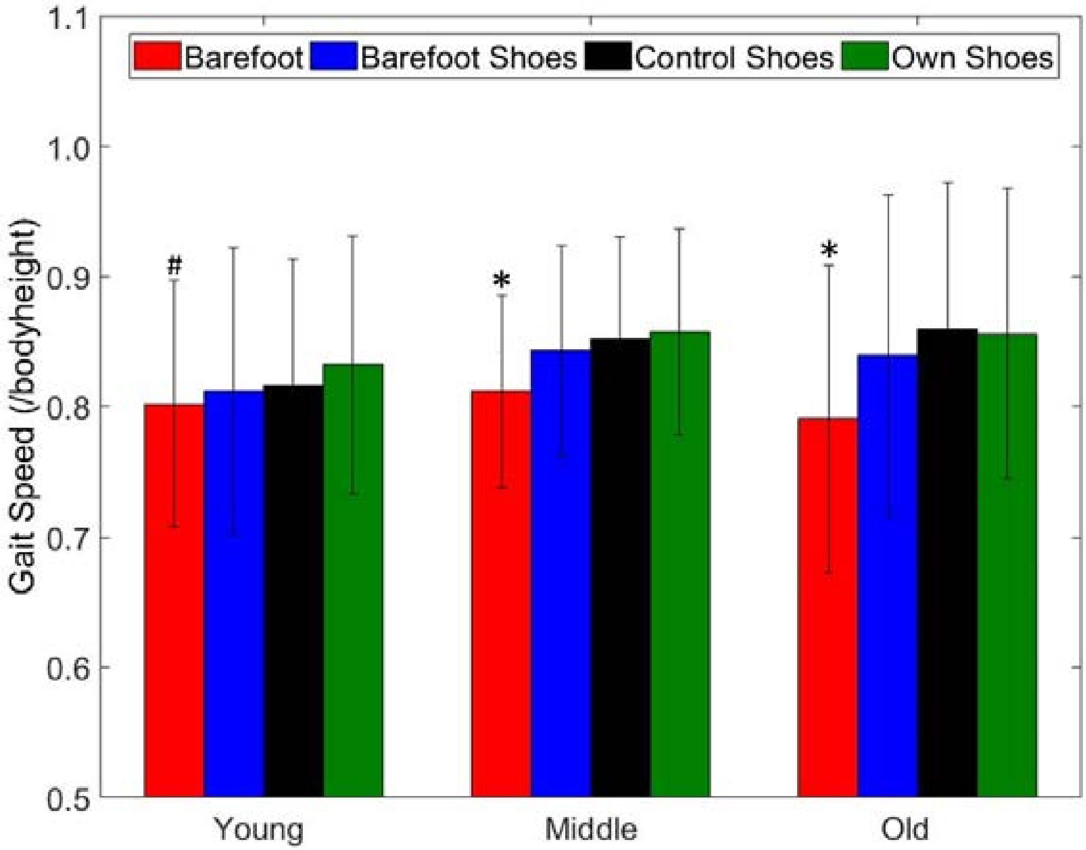

Figure 9 - A graph to display the difference in gait speed (normalised to bodyheight) across the different footwear conditions and between ages. Data displayed is of means with error bars depicting standard deviations. * indicates a significant difference compared to all other footwear (p<0.05). # indicates a significant footwear difference compared to the Participants' Own Shoe (p<0.05).

Stride Length

Stride length, normalised to body height, revealed a statistically significant main effect of footwear ($F(3,201)=247.997$; $p<0.001$; $\eta^2_{partial}=0.787$), a main effect of age ($F(2,67)=3.517$; $p=0.035$; $\eta^2_{partial}=0.095$) and also a significant *footwear x age interaction* ($F(6,201) = 4.311$; $p=0.001$; $\eta^2_{partial} = 0.114$). All age groups had a shorter stride length when BF compared to all other footwear (Young: 3.64% shorter than BLF, 5.70% shorter than CS, 6.54% shorter than OS; Middle: 4.46% shorter than BLF, 7.36% shorter than CS, 7.84% shorter than OS; Old: 6.90% shorter than BLF, 10.42% shorter than CS, 9.51% shorter than OS) and a shorter stride length in BLF than the conventional footwear (CS and OS) (Young: 1.99% shorter than CS, 2.81% shorter than OS; Middle 2.77% shorter than CS, 3.23% shorter than OS; Old 3.29% shorter than CS, 2.44% shorter than OS) (

Figure 1010). When walking BF the old age group had a shorter stride length than their young and middle age counterparts (6.91% shorter than Young and 7.36% shorter than Middle) and in their OS had a shorter stride length than the middle age group (5.93% shorter).

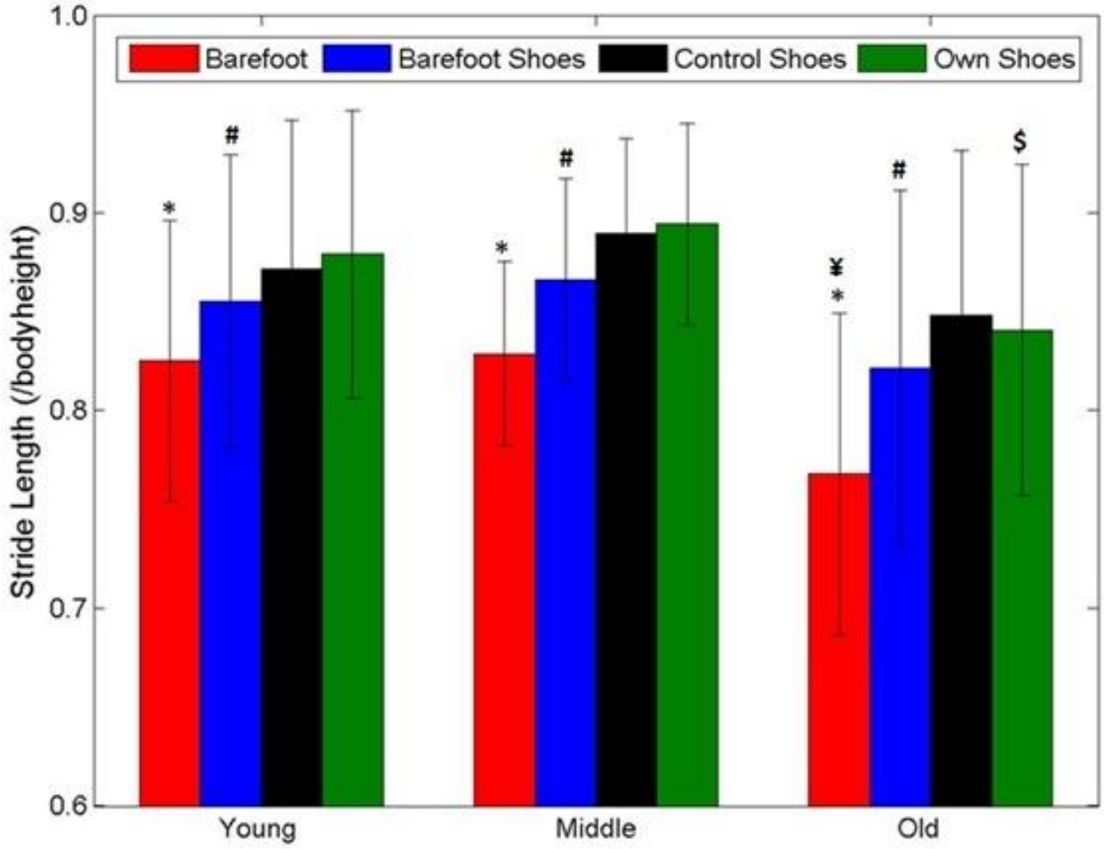

Figure 10 - A graph to display the difference in stride length (normalised to bodyheight) across the different footwear conditions and between ages. Data displayed is of means with error bars depicting standard deviations. * indicates a significant footwear main effect difference compared to all other footwear (p<0.05). # indicates a significant footwear main effect difference compared to the Control Shoe and the Participants' Own Shoe (p<0.05). ¥ indicates a significant footwear x age interaction difference to the Young and Middle age groups (p<0.05). $ indicates a significant footwear x age interaction difference to the Middle age group (p<0.05).

Ankle Angle at Contact (relative)

There was a statistically significant main effect of footwear ($F_{1.805, 120.930}$)=3.248; p=0.047; $\eta^2_{partial}$=0.046). Walking BF yielded a greater ankle angle at contact compared with walking in BLF (0.68°±0.22 greater when BF) (Figure 11).

Ankle Angle at Contact (absolute)

There was a statistically significant main effect of footwear ($F_{1.798, 120.471}$)=36.155; p=<0.001; $\eta^2_{partial}$ =0.350). Walking BF yielded a greater ankle angle at contact compared with walking in footwear (BLF: 1.12°±0.22 greater when BF, CS: 2.43°±0.28 greater when BF, OS: 2.70°±0.42 greater when BF). Walking in BLF resulted in greater ankle plantar flexion than conventional footwear (CS: 1.31°±0.20 greater when in BLF, OS: 1.58°±0.31 greater when in BLF).

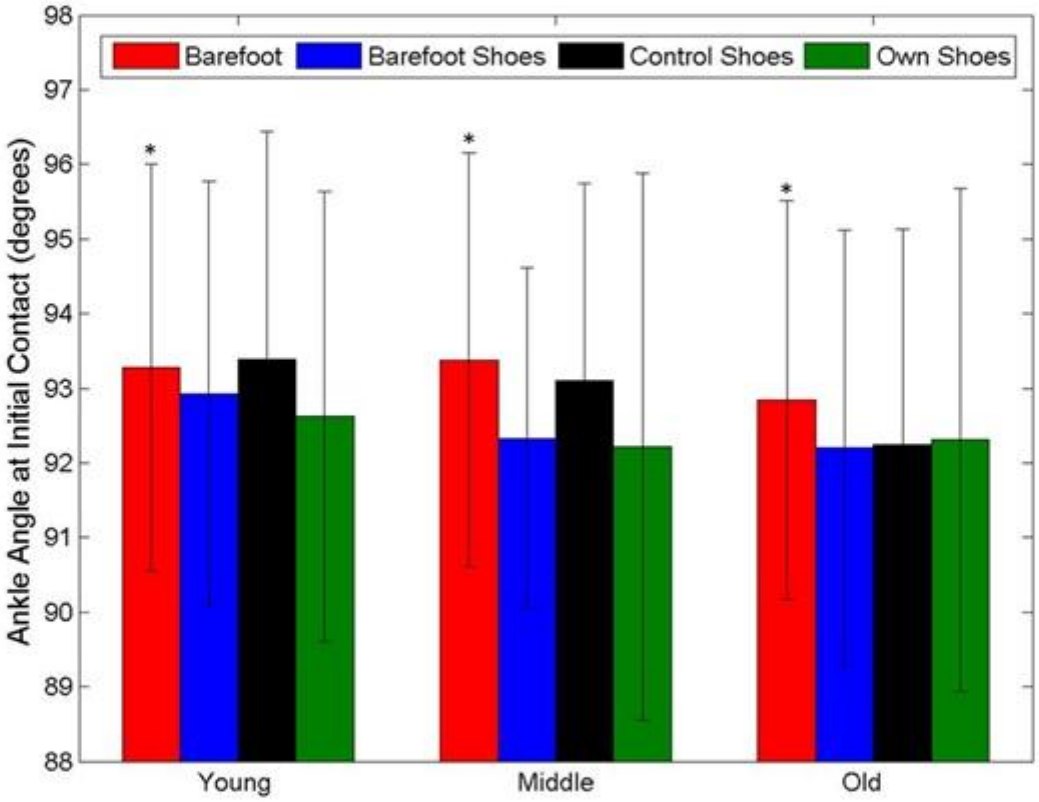

Figure 11 - A graph to display relative ankle angle (foot segment in relation to shank segment) at initial ground contact across the different footwear conditions and between ages. Data is displayed in means with error bars indicating standard deviations. * indicates a significant footwear main effect difference compared to the Barefoot Shoes condition (p<0.05).

Knee Angle at Contact

There was a statistically significant main effect of footwear ($F(2.315,155.084)$= 6.721;p=0.001; $\eta^2_{partial}$ =0.091). Walking in BLF increased the knee flexion at contact when compared to conventional footwear (CS: 1.06°±0.22 greater when in BLF, OS: 1.16°±0.32 greater when in BLF) (Figure 122).

Kinetics

Peak Loading Ground Reaction Force (GRF)

There was a statistically significant *main effect of footwear* ($F(2.496,167.211) = 13.287$; $p<0.001$; $\eta^2_{partial} = 0.165$). Walking BF resulted in a reduced peak loading GRF compared to all other footwear (BLF: 3.59% reduced when BF, CS: 3.29% reduced when BF, OS: 4.15% reduced when BF).

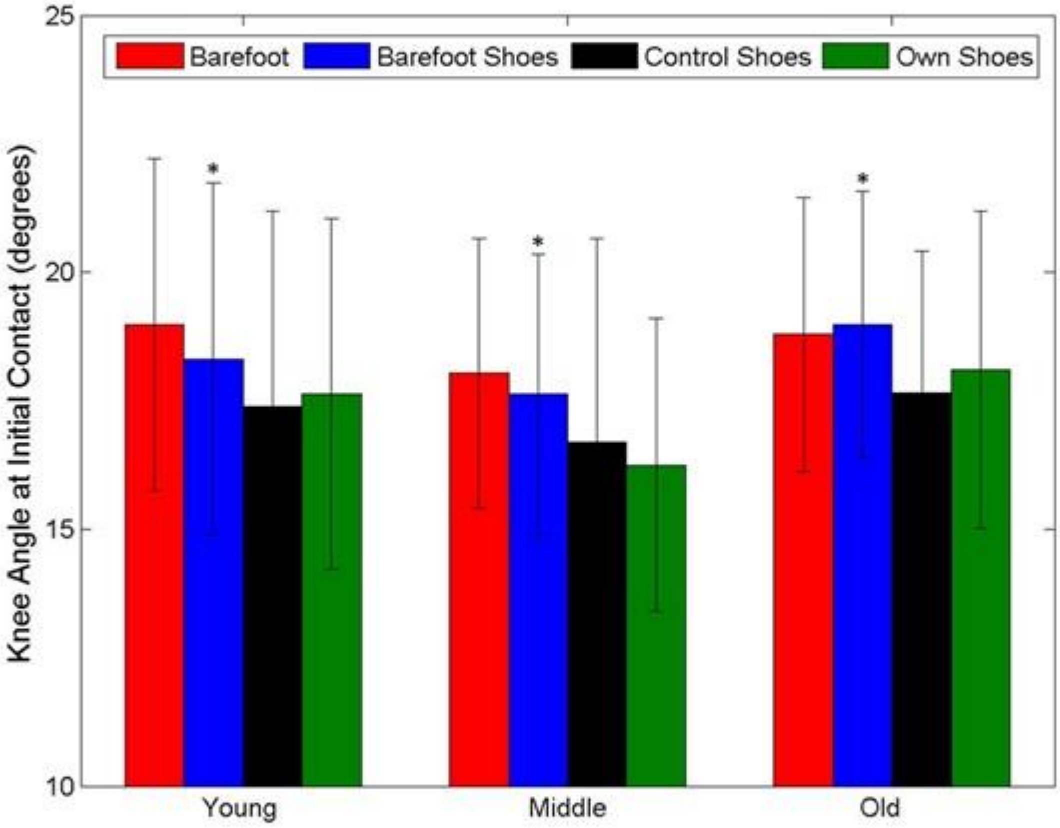

Figure 12 - A graph to display the knee flexion angle (shank segment with respect to thigh segment) at initial ground contact between the different footwear conditions and across age groups. Data is displayed as means with error bars indicating standard deviations. * indicates a significant footwear main effect difference compared to the Control Shoes and the Participant's Own Shoes conditions (p<0.05).

Discussion

This study investigated if kinematic and kinetic differences could be observed between walking BF, in BLF and conventional footwear across a range of ages. The first question concerned whether walking BF or in BLF results in similar kinematic and kinetic changes compared to conventional footwear. Although there were differences in the absolute amount of change, walking BF or in BLF showed the same trend of kinematic alterations including reduced step lengths, increased ankle plantar flexion and increased knee flexion at contact. In contrast, there was a reduction in peak loading GRF when BF but no reduction observed when wearing BLF compared to conventional footwear.

The second question concerned if the kinematic and kinetic changes witnessed when walking BF or in BLF are consistent across age. Although there are absolute differences across age groups (primarily the old age group showing differences to the middle and young) the response to walking BF or in BLF appears to be similar across age. All age groups reduced their stride length when walking BF or in BLF, showed similar changes in the ankle and knee angles and had a reduced peak loading GRF when walking BF.

Firstly it is important to note the difference in walking speed between the footwear conditions and acknowledging how the results should be interpreted as a result. In this study the walking speed was self-selected by the participant in order to ensure a natural walking pattern within each of the footwear conditions. It was found that with increasing age people walked slower barefoot than they did when wearing footwear within the range of 3% for the young up to 8% for the old. It has been shown previously that walking speed has an influence on kinematic correlates of walking including stride length and joint angle parameters (Caravaggi, Leardini et al. 2010, Chung and Wang 2010) and as such the reported changes in these parameters could be somewhat explained by the reduction in walking speed across conditions. That being said other recent studies investigating the effect of footwear have still seen differences in these kinematic parameters even when speed was controlled for highlighting that the footwear itself still provokes kinematic changes. In previous studies on running where the running velocity was kept constant across footwear conditions running barefoot resulted in a reduction in ankle dorsiflexion at contact as well as a reduction in step length compared to running in footwear (Hollander, Argubi-Wollesen et al. 2015). In a separate study where again the speed was kept constant but the footwear was changed it was found that there were differences in foot contact angle and spatio temporal parameters when wearing different types of minimalist footwear (Squadrone, Rodano et al. 2015). In support of this, in this study there were differences in the stride length and absolute ankle angle measures between the BLF and conventional footwear besides there being no significant difference in gait speed suggesting that walking speed cannot explain all of the differences witnessed. As such the effects of speed in this study should therefore be acknowledged however, noting the aforementioned explanations and small associated effect size of the gait speed difference ($\eta^2_{partial}$ = 0.125); the effects of footwear itself on kinematic changes should be recognised. Nevertheless it would be interesting to observe in future studies if speed was controlled for, how participants would increase their walking speed when barefoot, whether through increased stride rate and maintaining their shorter stride length and more plantarflexed foot strike or by increasing their stride length thus potentially affecting their foot strike angle in the process

In line with previous research there is a reduction in the stride length when walking BF. BLF also elicited reductions in stride length compared to conventional footwear but not to the same extent. This seems to suggest wearing something on your feet regardless of weight, degree of supportiveness or thickness of the sole causes gait alterations compared to BF gait. As BLF is unstructured, light and flexible the only substantial effect to the foot compared to unshod will be on the cutaneous afferents. The human foot has 104 cutaneous receptors on the sole of the foot and these are involved in movement control and maintaining balance (Kennedy and Inglis 2002). Previous non-human research has indicated reducing cutaneous afferent activity alters the locomotor cycle (Varejao and Filipe 2007); impaired walking pattern and stability is also witnessed in humans following plantar desensitisation (Lin and Yang 2011). It could be suggested that when wearing footwear the cutaneous afferents are under constant stimulation by the shoe material moving in contact with the skin. Particularly at the foot-ground contact phase of gait, movement of the foot within the shoe could lead to interference with stimulation applied across the whole foot.

Cutaneous afferent stimulation has also been shown to directly modulate lower leg muscle activity resulting in functional phase and location specific changes at the foot (Bent and Lowrey 2013, Zehr, Nakajima et al. 2014). Stimulation at the heel at initial stance brought about an increase in the medial gastrocnemius (MG) activity to produce plantar flexion whereas in contrast, stimulation at the forefoot at initial stance had the opposite effect and a suppression of MG activity (Zehr, Nakajima et al. 2014). This demonstrates the ability of the foot sole afferents to provide pressure information leading to muscle activity changes to modulate limb loading and foot placement and highlights the role of cutaneous afferents in balance control during locomotion. Clearly if the cutaneous afferents are being stimulated by the shoe structure across the whole foot sole and on the dorsal aspect of the foot, this may be interfering with the cutaneous sensory system inducing kinematic changes and could explain why walking in BLF is different to BF.

Overall BLF could yield the most effective gait by not entirely mimicking the kinematics of walking. Previous studies have suggested that independent of one another slower walking speed and longer step lengths are associated with an increased fall risk (Cromwell and Newton 2004, Bhatt, Wening et al. 2005, Moyer, Chambers et al. 2006, Espy, Yang et al. 2010). In the present study this reduction in gait speed and step length was observed when BF. Previous research states reducing step length serves to offset the decline in gait speed to ameliorate any increase in the risk of a slip-induced fall (Espy, Yang et al. 2010). Interestingly our results show that gait speed was preserved when wearing BLF even though step lengths were reduced. Therefore wearing BLF could offer the best alternative.

Knee and ankle joint differences at the point of foot-ground contact were also witnessed with respect to footwear. Greater plantar flexion was witnessed when BF compared to BLF when the ankle angle was computed in relative terms. Likewise when in BLF there was significantly greater relative knee flexion at initial contact compared to conventional footwear. This increased relative knee flexion would lead to a decrease in the absolute ankle plantar flexion required to produce a change in foot angle. In support of this, when the ankle angle was calculated in absolute terms, BF showed greater ankle plantar flexion compared to all other footwear but also BLF exhibited greater ankle plantar flexion than the conventional footwear. This illustrates how changing footwear doesn't just influence the foot segment, which has been directly affected, but also evokes changes further up the kinematic chain.

Kinetic data revealed a significant reduction in the peak loading GRF when BF but not in BLF. It is well known that the medial longitudinal arch of the foot, supported by the plantar aponeurosis (Carlson, Fleming et al. 2000) and active stiffening via intrinsic muscles (Kelly, Cresswell et al. 2014), helps control transmission of load (Erdemir, Hamel et al. 2004). Previous research has explained that the flatter foot at contact exhibited when BF enables the foot to take advantage of this mechanism to greater effect and explains the decrease in loading GRF (Lieberman, Venkadesan et al. 2010, Perl, Daoud et al. 2012). However, walking in BLF also lead to a significant increase in plantar flexion but no corresponding reduction in peak loading GRF was witnessed. The degree of plantar flexion increase from conventional footwear was less in BLF than seen BF which could imply there is a boundary which needs to be reached to effectively reduce the loading GRF.

De Clerq and colleagues have shown in their work on strike angle differences during running how small changes in strike angle can have a significant effect on impact modulation (Breine, Malcolm et al. 2016, De Clercq, Breine et al. 2017). They identified that a typical rear-foot strike where the contact occurs at the heel followed by ankle plantarflexion exhibited similar impact reducing outcomes to that of a mid-foot or fore-foot strike where the ankle undergoes dorsiflexion following initial contact (Breine, Malcolm et al. 2016). In contrast they identified that an atypical rear foot strike whereby the foot strike angle was effectively flat due to a fast anterior displacement of the COP from the initial contact at the rearfoot to early first metatarsal contact had impaired impact reducing capacity (De Clercq, Breine et al. 2017). This highlights how the foot position at initial contact and allowing for ankle plantarflexion or dorsiflexion to occur is inexplicably linked to impact modulation and the transmission of load (Breine, Malcolm et al. 2016).

The differences in the effect of kinematics on loading forces between BLF and BF conditions could also be attributed to a psychological factor, specifically the perception of protection. This theory was promoted when investigating the potential hazard of deceptive advertising (Robbins and Waked 1997). When participants were led to believe the surface they were stepping on to offered superior protection through the best cushioning material, they experienced the greatest impact force compared to no cushioning and when they perceived the surface to have lesser cushioning properties. In reality, the surface didn't change but it highlighted how activity could be changed based on perceived protection and this could be applied to our results. As participants are aware they have something on their feet they may perceive they are protected. In contrast, when BF they may assume they have to take greater care with foot placement as there is nothing under the foot to offer protection and thus a more cautious gait adopted.

Another contributing factor could be the decreases in speed and stride lengths observed when walking BF. Contacting the ground closer to the vertical projection of the body's center of mass and with reduced prior horizontal velocity could contribute to this peak loading GRF reduction. Correlation analysis showed that gait speed was responsible for approximately 20-25% and stride length for approximately 13-17% of the peak loading GRF across footwear conditions. It's reasonable to suggest a combination of these factors along with changes in the joint angles and improved uptake of load by lower leg and foot musculature when walking BF account for this attenuation of the loading GRF.

All age groups respond the same way when exposed to walking BF or in BLF regardless of overall age differences. All ages experienced a reduction in step and stride length, a reduced peak GRF, greater ankle angle when BF and greater knee flexion angle when in BLF. A potential confounding reason for this lack of differences across age could be our decision to ensure the participants were closely matched across age in terms of walking ability. All participants displayed a good level of overall mobility and a low occurrence of falls and often these occurred during activities such as hiking (Table 6). It's therefore possible that, particularly those in the older age groups, comprise the more physically active amongst the population. This is a typical bias of many studies on the elderly population due to the increased difficulty of reaching and motivating non-active elderly to participate in a locomotion experiment compared to their active peers. Future study may be warranted into the effects of walking BF or in BLF on those less physically active and potentially at an increased risk of falling.

Another avenue for future research could be to explore if there is an adaptation effect when going from walking in conventional footwear to walking barefoot or in minimalist footwear. The cross-sectional nature of this study shows how participants react to these footwear conditions acutely but it would be interesting to see over a longer period of time spent walking in these footwear conditions if these changes persist or if a degree of kinematic adaptation occurs. Increasing the number of trials or recording samples within a longer walking timeframe could be a method of assessing this within a cross-sectional study design or alternatively including multiple assessment time points within a longitudinal footwear intervention study design would be useful to determine if any transition or adaptation period to this footwear type is witnessed.

Conclusion

In summary, this study investigated the kinematic and kinetic differences between walking BF, in BLF and conventional footwear and if there were changes with respect to age. Clear kinematic and kinetic differences were observed between walking BF and in footwear. These include increased ankle plantar flexion and knee flexion at foot-ground contact, a reduction in step and stride length, a reduction in gait speed and a reduced peak loading GRF when BF. Wearing BLF is not equivalent to walking BF but the differences in kinematics are reduced. Wearing BLF may result in a more stable gait with reduced step lengths whilst preserving speed. Wearing footwear causes changes further up the kinematic chain and not solely at the foot segment. The changes witnessed between BF and BLF footwear may be the result of cutaneous afferent interference and the perception of protection when wearing any type of footwear. The response to walking BF or in BLF is consistent across age in fit and healthy individuals.

Chapter 4

Does walking barefooted or in minimalist footwear increase lower leg muscle activation?

Simon Franklin, François-Xavier Li, Michael J. Grey

Abstract

Ageing is associated with a decline in muscle strength and impaired sensory mechanisms which contribute to an increased risk of falls. Walking barefooted has been suggested to promote increased muscle strength and improved proprioceptive sensibility through better activation of foot and ankle musculature. Minimalist footwear has been marketed as a method of reaping the suggested benefits of barefoot walking whilst still providing a protective surface. The aim of this study was to investigate if walking barefoot or in minimalist footwear provokes increased muscle activation compared to walking in conventional footwear. Seventy healthy adults (age range 20-87) volunteered for this study. All participants walked along a 7m walking lane five times in four different footwear conditions (barefoot (BF), minimalist shoes (MSH), their own shoes (SH) and control shoes (CON)). Muscle activity of their tibialis anterior (TA), gastrocnemius medialis (GCM) and peroneus longus (PL) were recorded simultaneously and normalised to the BF condition. MSH are intermediate in terms of ankle kinematics and muscle activation patterns. Walking BF or in MSH results in a decrease in TA activity at initial stance due to a flatter foot at contact in comparison to conventional footwear. Walking BF reduces PL activity at initial stance in the young and middle age but not the old. Walking in supportive footwear appears to reduce the balance modulation role of the GCM in the young and middle age but not the old, possibly as a result of slower walking speed when BF.

Introduction

As proponents of bipedal locomotion, humans possess an inherently unstable system requiring constant modulation by balance mechanisms in order to prevent falling (Winter 1995). For millions of years humans walked barefoot (BF) and as such, the feet have evolved to cope with the demands of bipedal locomotion. The human foot comprises 104 cutaneous mechanoreceptors responsible for sensing changes in pressure, vibration and skin stretch and their distribution across the foot highlight their role in balance and movement control (Kennedy and Inglis 2002). These plantar mechanoreceptors contribute to the automatic modulation of the phases of gait via the detection of pressure changes during foot-ground contact (Fallon, Bent et al. 2005) and along with proprioceptive afferents assist in the planning and correction of movement (Kafa 2015). This information is essential for controlling static and dynamic stability.

Footwear habits have since changed and there is suggestion that modern day footwear may be impairing the capability of these afferent receptors. Wearing highly structured and supportive shoes could be limiting the input as the foot is not as susceptible to changes in shape, pressure and touch due to the confines placed upon it. This idea has been furthered by Nigg (2015), who hypothesised that walking BF can activate the smaller muscles within the feet as well as the larger muscles crossing the ankle joint. He suggested that activating these smaller muscles might provide greater stability as they can more quickly sense changes in different directions and with smaller amounts of force being required (Nigg 2009). Whilst this position paper was primarily focussed on running performance and injury reduction, the premise of improved stability by activating the smaller muscles, could have implications for the older population in terms of fall prevention.

Wearing footwear may also lead to foot muscle weakening due to the reduction in the stresses put upon the foot by means of supportive features within modern day shoes (Lieberman 2012).

It is well known that ageing causes a decline in muscle strength along with sensory impairments and these factors contribute to the increased susceptibility to falls. Research has shown that wearing shoes with a flexible segmented midsole (Nike Free 3.0) for athletic training resulted in a significant increase in toe flexor strength (Goldmann, Potthast et al. 2013). This suggests that changing the footwear worn to less supportive and more flexible shoes, potentially allowing for greater movement within the foot may better activate the foot muscles. As such, research is now required to determine if walking in minimalist shoes (MSH), footwear comprised of minimal structure and high flexibility, better activates afferent and efferent mechanisms and if this can have a positive influence on stability.

A recent systematic review investigated the effect of footwear, or the lack of footwear, on walking (Franklin, Grey et al. 2015). Aside from outlining the overall kinematic differences between shod and barefoot walking, the review highlighted the paucity of research on BF and MSH use in older age populations as well as the distinct lack of study on muscle activity differences between shod, minimally shod and unshod conditions. Previous study has shown that there are significant differences in muscle activation during walking between the young and the old. Older adults have a significantly greater magnitude of agonist and antagonist muscle co-activation and this corresponds to a 19.2% increase in their energy cost of walking on level ground (Hortobagyi, Finch et al. 2011). It has also been shown that older adults tend to have greater muscle co-activation prior to impact in order to prepare for load absorption by attempting to increase joint stiffness compensating for the reduction in overall strength (Hsu, Wei et al. 2007). No studies to date have investigated the effect of footwear but by walking barefoot it could be that similar increases in muscle activity may be witnessed across the age groups in order to increase joint stability to compensate for the loss of support provided by the footwear. Furthermore with greater years spent in conventional footwear, older adults may have become more reliant on the supportive aspects of their footwear and thus walking barefoot may evoke greater increases in muscle co-activation than seen in the younger adults.

Consequently, the aims of this study were to investigate if walking BF or in MSH share the same lower leg muscle activation patterns and to determine if greater muscle activation is provoked compared to conventional footwear. We also aimed to determine if there were any differences with respect to age and years spent wearing structured footwear. We hypothesised that muscle activation patterns between walking BF and in MSH would be similar and that there would be greater activation of the lower leg muscles during the stance phase in these conditions. We also hypothesised that the old age group would show a greater increase in muscle activity when walking BF compared to wearing structured footwear.

Methods

70 healthy adults (27 males, age range 20-87years) participated and were split into 3 age groups (YOUNG <40 years (n=20), MID >40 years and <70 years (n=30) and OLD >70 years (n=20) (Table 7). All participants were able to ambulate independently and had no known gait disorders or abnormalities. All participants completed a general health questionnaire and signed an informed consent prior to testing as approved by the University ethics committee (ERN_14-0560).

Kinematic markers were placed bilaterally at the lateral epicondyle (R/LKNE), base of the calcaneus (R/LHEE), medial mallelous (R/LANK) and first metatarsophalangeal joint (R/LTOE). When wearing footwear, markers were attached to the shoes in the same positions as determined by palpation. Surface EMG electrodes (Wave Wireless EMG, Cometa Systems, Milan) were placed on the right leg over the belly of the tibialis anterior (TA), peroneus longus (PL) and gastrocnemius medialis (GCM) muscles in the positions outlined by the SENIAM guidelines (Hermens, Freriks et al. 2000).

Figure 13 - A) The control shoe worn by males (Style Code: 50109, Hobos Mens). B) The control shoe worn by females (Style Code: 10001, Hobos Womens). C) and D) The unisex minimalist shoe worn by all participants (Product ID: 2169, Two Barefeet Boarding Co.).

At each site the skin was shaved, abraded and cleaned with an alcohol swab before attaching two disposable pre-hypoallergenic gelled (1cm diameter) self-adhesive Ag/AgCl snap electrodes with an inter-electrode distance of 1.5cm. The EMG signals were collected at a rate of 2000Hz, amplified with a gain of 1000 (input impedance 20MΩ, CMRR >100dB), and bandpass filtered from 10–1000Hz (De Luca, Gilmore et al. 2010).

Thirteen Vicon MX cameras (Vicon, Oxford, UK) recording at a sampling rate of 250Hz collected three dimensional kinematic data. Gait cycle phases were computed using the R/LHEE and R/LTOE markers and absolute ankle angle was determined using the foot vector (RANK and RTOE markers) with respect to a vertical vector from the ankle.

Participants walked at a self-selected speed through a 7m walking lane from a mark based on 3 practice trials such that 3 steps were taken prior to the point of data collection commencing. Participants completed 5 trials in each of the four randomly assigned footwear conditions. The footwear were BF, a MSH (Product ID: 2169, Two Barefeet Boarding Co.), a control shoe (CON)

(Style Code: 10001, Hobos Womens, Style Code: 50109, Hobos Mens) and the participants own footwear (SH). The mens and womens control shoes were different so as to be specific to gender sizes and fit appropriately to their different foot shapes. Both footwear however had similar features conforming to that of a recommended conventional shoe including a supported heel-collar, low bevelled heel, fastening mechanism and a thin firm midsole (Menant, Steele et al. 2008). EMG and kinematics were recorded in synchrony.

Post-processing of the data was completed using custom-written scripts in Matlab (MATLAB, The MathWorks, Natick, MA, USA). Kinematic data were low pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12Hz. Muscle activity data were zero offset, before being full wave rectified and then low pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 10Hz. Once frequency matched to the synchronised kinematic data, the linear envelope for each participant's trials were cut from right heel strike to right heel strike and normalised to the gait cycle (0-100%). Each trial comprised 1-4 recorded cycles. Maximal voluntary contractions were not completed due to previous reports of poor reliability in achieving a maximum for the PL (Ozaki, Mizuno et al. 1999, Hagen, Lahner et al. 2015); therefore all cycles for each participant were collated and normalised to the average of all the cycles when BF. The normalised cycles within each respective trial were ensemble averaged to provide an average muscle activity trace for each trial. Each trial was then divided into stance and swing phases and the stance phase sub-divided into Initial Double Support (IDS), Single Support (SS) and Late Double Support (LDS). The mean activity was then computed for each muscle within each gait cycle phase. Due to recording errors in certain trials resulting in missing data the number of trials available for comparison was limited to 4.

Mixed design repeated measures Analysis of Variance (ANOVA) were completed for each variable to determine the differences across footwear (BF vs MSH vs CON vs SH), trial (1:4) and age group (YOUNG vs MID vs OLD). Mauchly's test of sphericity was completed to ensure validity and in the case where this test was violated a Greenhouse-Geisser correction was applied. Statistical analyses were performed using SPSS V.22 for Windows (IBM Corporation, Somers, NY) with levels of significance set to $p<0.05$.

Results

Age Group	No. In group	Age (years)	BMI (kg/m^2)	Own Shoe weight (g)
Young	20	27.85 (4.83)	23.25 (3.46)	293 (70.29)
Middle	30	54.85 (9.85)	25.04 (3.48)	319 (114.38)
Old	20	77.55 (4.39)	25.21 (4.39)	304.25 (71.6)

Table 7– A summary of the participant's statistics. Data is displayed as means (S.D). BMI = Body Mass Index; S.D = Standard Deviation; g = grams

Tibialis Anterior

In the stance phase there was a significant effect of footwear ($F(2.657,177.989)=23.920$, $p<0.001$, $\eta_p^2=0.263$), a significant effect of trial ($F(3,201)=4.643$, $p=0.004$, $\eta_p^2=0.065$) but no significant effects of age. Walking BF exhibited lower TA activation compared to MSH, CON and SH by 0.096mV±0.032, 0.249mV±0.038 and 0.242mV±0.039 respectively. Walking in the MSH showed lower TA activation than the CON and SH conditions by 0.153mV±0.035 and 0.146mV±0.036 respectively. There was greater TA activation in trial 1 than trial 3 and 4 by .063mV±0.022 and 0.056mV±0.020 respectively. After stance phase subdivision there was a significant effect of footwear in the IDS phase ($F (2.190,146.759) = 37.416$, $p<0.001$, $\eta_p^2=0.358$) and SS phase ($F (3, 201) = 20.145$, $p<0.001$, $\eta_p^2=0.231$) but not in the LDS phase but no significant effects of age or trial. In the IDS phase walking BF lead to lower TA activation in the IDS phase compared to the MSH, CON and SH conditions by 0.238mV±0.037, 0.547mV±0.061 and 0.489mV±0.071 respectively whilst walking in MSH showed lower TA activation than the CON and SH conditions by 0.309mV±0.056 and 0.251mV±0.064 respectively. In the SS phase walking

BF resulted in reduced TA activation during the SS phase compared to the MSH, CON and SH conditions by 0.290mV±0.059, 0.338mV±0.054 and 0.351mV±0.053 respectively.

Gastrocnemius Medialis

In the stance phase there was a significant interaction effect of footwear*age (F(4.674,156.570)=3.175, p=0.011, η_p^2=0.087) but no significant effects of trial. The YOUNG showed lower GCM activation when wearing CON compared to BF, the MSH and SH conditions by 0.161mV±0.029, 0.194mV±0.039 and 0.108mV±0.031 respectively; the MID had a lower GCM activation when wearing the CON compared to MSH and SH conditions by 0.130mV±0.032 and 0.079mV±0.026 respectively whereas the OLD showed no differences across footwear. Stance phase subdivision displayed a significant footwear*age interaction effect in the SS phase (F (4.814,161.253) = 3.085, p=0.012, η_p^2=0 .084) and a significant main effect of footwear in the LDS phase (F (2.198,147.276) = 14.169, p<0.001, η_p^2=0.175) but no significant differences in the IDS phase or any significant effects of trial. In the SS phase the YOUNG exhibited lower GCM activation when wearing the CON compared to BF, the MSH and SH conditions by 0.210mV±0.031, 0.141mV±0.037 and 0.113mV±0.027 respectively; the MID showed lower GCM activation in the CON compared to the MSH by 0.099mV±0.027 whereas the OLD showed no differences across footwear. Conversely in the LDS phase it was seen that walking BF leads to lower GCM activation during the LDS phase compared to the MSH, CON and SH conditions by 0.626mV±0.145, 0.975mV±0.177 and 1.260mV±0.257 respectively.

Peroneus Longus

In the stance phase there was a significant main effect of footwear (F(2.328, 155.946)=5.335, p=0.004, η_p^2=0.074) and a trial*age interaction (F (3.859, 129.281) = 2.815, p=.030, η_p^2= .078). Walking BF lead to reduced PL activation compared to the CON and SH conditions by 0.067mV±0.023 and 0.124mV±0.034 respectively; the YOUNG had greater PL activity in the 2nd

trial compared to the 4th by .107mV±.028 and the OLD had greater PL activation in the 1st trial compared to the 3rd by .128mV±.039. With stance phase subdivision there was a significant interaction effect between footwear*age (F (5.045, 2.805) = 2.805, p=0.018, η_p^2 =0.077) in the IDS phase, a significant main effect of footwear in the LDS (F (3, 201) = 5.414, p=0.001, η_p^2 =0.075) but no significant differences in the SS phase or any significant effects of trial. In the IDS phase the YOUNG had a reduced PL activity when BF compared to the CON and SH conditions by 0.368mV±0.086 and 0.313m±0.088 respectively and also when wearing the MSH compared to CON and SH conditions by 0.278mV±0.082 and 0.223mV±0.075 respectively. The MID displayed reduced PL activity when BF compared to the MSH, CON and SH conditions by 0.153mV±0.051, 0.366mV±0.070 and 0.390mV±0.072 respectively and when wearing the MSH compared to CON and SH conditions by 0.213mV±0.067 and 0.237mV±0.061 respectively whilst the OLD showed no differences between footwear. In the LDS phase walking BF lead to lower PL activation compared to the CON and SH conditions by 0.222mV±0.077 and 0.238mV±0.085 respectively.

Ankle Angle Heel Strike

There was a significant effect of footwear (F(2.484,166.422)=64.094, p<0.001, η_p^2 =0.489) and a significant interaction effect of trial*age (F(3,201)=2.562, p=0.038, η_p^2 =0.037). Walking BF resulted in greater plantar flexion compared to the MSH, CON and SH conditions by 3.118°±0.385, 5.597 °±0.487 and 5.866°±0.599 respectively. Walking in the MSH resulted in greater plantar flexion compared to the CON and SH conditions by 2.480°±0.405 and 2.748°±0.502 respectively. The OLD had greater plantarflexion in trial 2 compared to trial 3 by 0.595°±0.209.

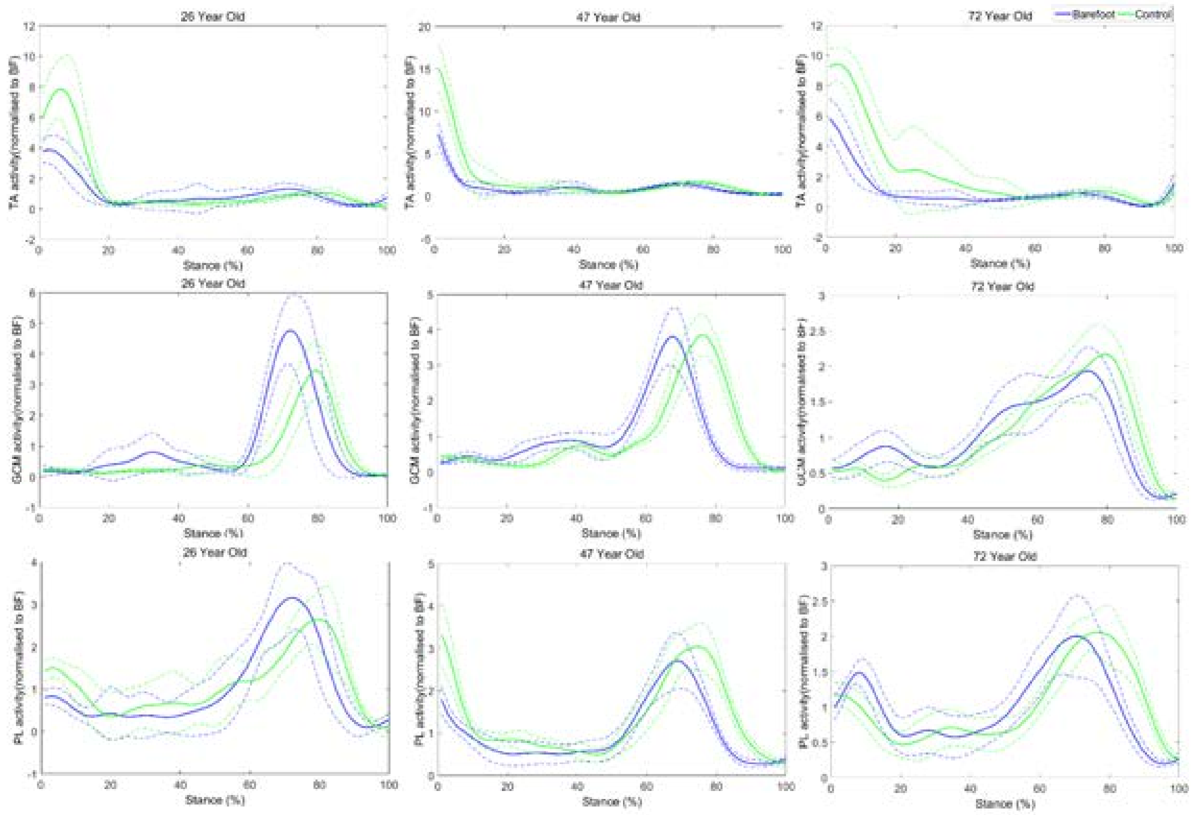

Figure 14 - Graphs to illustrate the average activity of the tibialis anterior (TA), gastrocnemius medialis (GCM) and peroneus longus (PL) in the barefoot (blue lines) and control shoe (green lines) conditions after each cycle was normalised to the average activity of all cycles within the stance phase during the barefoot (BF) condition for each participant. Dotted lines indicate the standard deviation across all the cycles within each respective footwear condition. The left graph is a representative participant from the young age group (26years old), the middle graph a representative participant from the middle age group (47years old) and the right graph a representative participant from the old age group (72years old).

Gait Speed

There was a significant interaction effect between footwear*age (F(6,201)=4.322, p=0.002, η_p^2 =0.114) and a significant interaction effect of trial*age (F(4.808,161.052)=2.815, p=0.020, η_p^2 =0.078). The YOUNG walked slower BF compared to when wearing the CON and SH conditions by 0.032m/sec±0.011 and 0.034m/sec±0.013 respectively. The MID walked slower BF than the MSH, CON and SH conditions by 0.038m/sec±0.008, 0.067m/sec±0.009 and 0.065m/sec±0.010 respectively and walked slower in the MSH than the CON conditions by 0.029m/sec±0.010. Similarly the OLD walked slower BF than the MSH, CON and SH conditions by 0.064m/sec±0.010, 0.108m/sec±0.011 and 0.101±0.013 respectively and walked slower in the MSH than the CON and SH conditions by 0.045m/sec±0.012 and 0.038±0.013 respectively. The YOUNG also walked slower in trial 4 than trial 3 by 0.011m/sec±0.004.

Discussion

This study was designed to determine if there are lower leg muscle activity differences between walking barefoot, in minimalist shoes or conventional footwear (CON and SH). The results illustrate that the first hypothesis is to be rejected as the degree of muscle activation differed between BF and MSH conditions. Contrary to our second hypothesis, walking BF or in MSH was not observed to lead to increases in muscle activity during stance and in the TA and PL was seen to be lower than in conventional footwear. Furthermore the third hypothesis is also to be rejected as the OLD age group showed the least amount of differences across footwear conditions. There was no increase in stance phase lower leg muscle activity when walking BF or in the MSH condition. There was also no significant footwear*trial interaction effects suggesting that no adaptation to each footwear condition was witnessed and the responses to the footwear were consistent across trial number.

The GCM, has been attributed a role in balance control during gait due to its ability to modulate the vertical displacement of the centre of mass (CoM) in relation to the centre of pressure (CoP) thus acting to prevent falling (Honeine, Schieppati et al. 2013). During the SS phase the body pivots over the ankle and approaches the LDS phase. The CoM trajectory follows an arc shape whereby the top of the arc is the point where the CoM is directly above the ankle and after this point it begins to lower due to the separation between the CoM-CoP and influence of gravity. The GCM's role is to increase its activity in order to maintain vertical support and prevent the CoM trajectory dropping too low by increasing the anterior progression of the CoP (Francis, Lenz et al. 2013). This has an indirect effect on step length and gait velocity (Honeine, Schieppati et al. 2013). In this study it was observed that when walking in the CON shoe there was a decrease in

GCM activity compared to all other footwear conditions in the YOUNG; a decrease compared to the MSH and SH conditions in the MID but no difference across footwear in the OLD. The CON provides in-built support and greater overall anterior-posterior length due to a large sole size and therefore it was hypothesised that less emphasis would be placed on the GCM to control the CoM vertical displacement. This was only witnessed in the YOUNG with the MID showing no difference between the CON and BF conditions and the OLD showing no difference across all footwear conditions. A confounding factor which could partially explain these results could be the effect of walking speed. Consistent with previous findings (Franklin, Grey et al. 2015), all ages walked slower BF however the amount of discrepancy grew with increasing age such that the difference in speed between BF and the CON condition in the OLD was over 3 times greater than it was in the YOUNG. Walking slower decreases the balance modulation role of the GCM and therefore this may offset the increase in muscle activity due to the removal of supportive shoe structures potentially explaining the lack of difference witnessed in the OLD age group.

PL activity was reduced when participants walked BF compared with conventional footwear. As the PL plays a role in the maintenance of lateral stability around the foot during walking (Louwerens, van Linge et al. 1995), our data suggest that we are prone to greater lateral instability during the initial loading phase when wearing conventional footwear. This could be a result of reduced foot position awareness when wearing conventional footwear. It’s been suggested that when BF smaller intrinsic foot muscles are activated and these sense and control joint stability quickly and with little force being required as they are more sensitive to smaller changes in different directions than the larger muscles crossing the ankle (Nigg 2009). If this is the case, then these larger muscles would be required less for joint stabilisation and thus a reduced activation in these muscles may be observed. This was witnessed in the YOUNG and MID age groups however the OLD showed no differences across footwear. This could hint at age-related detriments in proprioceptive acuity. It’s been previously shown that older adults have an increased threshold to touch, pressure and vibration whilst joint position acuity is also negatively affected (Perry 2006, Relph and Herrington 2016). It could be that this insensitivity is prominent in the smaller intrinsic foot muscles and hence the increased afferent information available to the YOUNG and MID when minimally/unshod may not have the same benefit to the OLD. What is clear however is that removing supportive footwear in the OLD age group does not worsen their lateral stability as implied by the lack of difference in PL activity. Proprioceptive acuity has however been shown to be receptive to improvements with training in elderly women (Thompson, Mikesky et al. 2003). As this was only an acute exposure to walking barefoot it is not known whether consistent exposure to minimally/unshod conditions could promote proprioceptive improvements of the foot muscles leading in similar results to the younger age groups. Further study is required to investigate the activation patterns of these smaller muscles within the foot to explore this theory.

A decrease in TA activity during initial stance was also observed when walking BF and in MSH shoes compared to conventional footwear. Whilst the TA's primary role is to provide toe clearance during the swing-phase of the gait cycle (Barthélemy D, Grey MJ et al. 2011), it also assists in the control of stability during weight-acceptance at initial contact by eccentrically contracting to lower the foot to the ground. In the previous chapter it was observed that there was a reduction in GRF when walking barefoot and this was attributed to an increase in plantarflexion at initial contact allowing for a rotational moment about the ankle helping to dissipate the force as previously described (Breine, Malcolm et al. 2016). This corresponds to the decrease in TA activity witnessed here as this change in foot position at initial contact to a more plantarflexed position, potentially as a result of the decline in walking speed when minimally/unshod, requires reduced input from the TA to control the load and dissipate the force as previously mentioned (Lieberman, Venkadesan et al. 2010).

It should be stated that by maintaining shoe integrity and affixing markers to the shoe surface rather than through cut-outs, small discrepancies in marker position between BF and shod may be present. This could explain a small part of the strike angle differences. Additionally as there was a difference in the speed of walking between walking barefoot and walking in footwear across the age groups it should be noted that some of the differences witnessed in muscle activity could be due to the reduction in speed and decreased reliance on the lower limb musculature for impact modulation (Chung and Wang 2010). Whilst speed could have been controlled experimentally to eliminate this speed effect, in this study participants were instructed to self-select their walking speed in all footwear conditions to ensure natural gait characteristics in response to the different conditions. It is clear that a slower walking speed is a consequence of the barefoot condition likely for impact reducing purposes however previous research has also shown how if speed is kept constant, kinematic alterations still exist in response to different footwear types (Hollander, Argubi-Wollesen et al. 2015) likely still as an impact reducing strategy (Breine, Malcolm et al. 2016). As such it would be interesting for future research if the speed is controlled for and the pure effect of footwear on lower leg muscle activity can be examined.

Conclusion

In summary, this study investigated the muscle activity differences between walking BF, walking in MSH and conventional footwear. MSH are intermediate in terms of ankle kinematics and muscle activation patterns. Walking BF and in MSH results in a decrease in TA activity at initial stance due to a flatter foot at contact. Walking BF also leads to a reduction in PL activity at initial stance in the young and middle age but not the old. Walking in supportive footwear leads to a reduction in GCM activation in the young and middle age but not the old, possibly as a result of slower walking speed when BF.

Chapter 5

Assessing the reliability of a customised device at measuring great toe flexor muscle (gTFM) strength and the importance of its study.

Simon Franklin, François-Xavier Li, Michael J. Grey

Abstract

Toe flexor muscles (TFM) have been identified as being particularly important in human locomotion for overcoming the mechanical inefficiency caused by ankle plantarflexion, and the force-length relationship, during the push off phase of gait; dorsal flexion occurs at the metatarsal phalangeal joints (MPJ) at push off to help produce the required propulsive force (Goldmann and Bruggemann 2012). TFM have also been shown to be important for functional balance particularly for maintaining stability in reaching and leaning tasks (Endo, Ashton-Miller et al. 2002). Training of the TFM to assess if gait and functional balance can be improved with increased strength is therefore of interest. For such studies to occur, a device is required to accurately and repeatedly measure the strength of the TFM. This study therefore aims to address this and assesses the reliability of a customised device at measuring great toe flexor (gTFM) strength whilst further discussing the importance of TFM strength to balance and gait.

Introduction

Internal foot muscle strength and in particular the toe flexor muscles (TFM) have been shown to be important in human locomotion (Misu, Doi et al. 2014) as well as postural stability (Mickle, Munro et al. 2009). Various devices have thus been tested in order to assess TFM to further investigate its relationship with balance and gait as well as using these to monitor the effect of improved TFM strength on athletic performance (Mickle, Chambers et al. 2008, Goldmann, Sanno et al. 2012). Assessing TFM strength can have its challenges due to variation in foot dimensions making it difficult to build a device to assess overall TFM strength. A previous study found that due to variation in foot dimensions not all participants were able to position all their toes on their custom-built dynamometer (Goldmann and Bruggemann 2012). Previous research has demonstrated that around 60% of the total force applied during walking is through the great toe with only 5-13% under the lesser toes (Hughes, Clark et al. 1990). It has also been shown that the great toe is fundamental in terms of static and dynamic balance (Chou, Cheng et al. 2009, Tanaka, Hashimoto et al. 1996). There was a significant linear relationship observed between body sway and the pressure under the great toe suggesting that the strength of the toes is of importance for maintaining stability during a dynamic balance task (Tanaka, Hashimoto et al. 1996). With the majority of work shown to be done by the great toe in relation to the lesser toes and its clear relevance to balance and gait measures, study into improving great toe strength particularly in those populations at risk of falling would seem pertinent. It is therefore essential to have an accurate and reliable method with which the force of the great toe flexor muscles (gTFM) can be measured. This study aims to assess the accuracy and reliability of repeated measurements of gTFM strength using a custom built device in order to validate its use in future studies.

Methods

Thirty healthy young adults (15 male (24.47 (±3.56) years, 77.6 (±10.19) kg), 15 female (20.47 (±2.36) years, 61.2 (±6.62) kg)) volunteered and participated in the study. All participants had no lower limb or foot muscle problems or any medical conditions which may have affected their ability to produce maximal force with their gTFM. Participants' were informed of the protocol and signed an informed consent as approved by the University's ethical committee before their body weight and longitudinal foot length of both feet were measured and the data recorded. They were then screened for foot dominance through one leg standing and the 'write word' test as used in prior studies (Velotta, Weyer et al. 2011). We specified the dominant foot as being the leg used for stabilisation rather than mobilisation as explained in a previous review on limb dominance (Sadeghi, Allard et al. 2000).

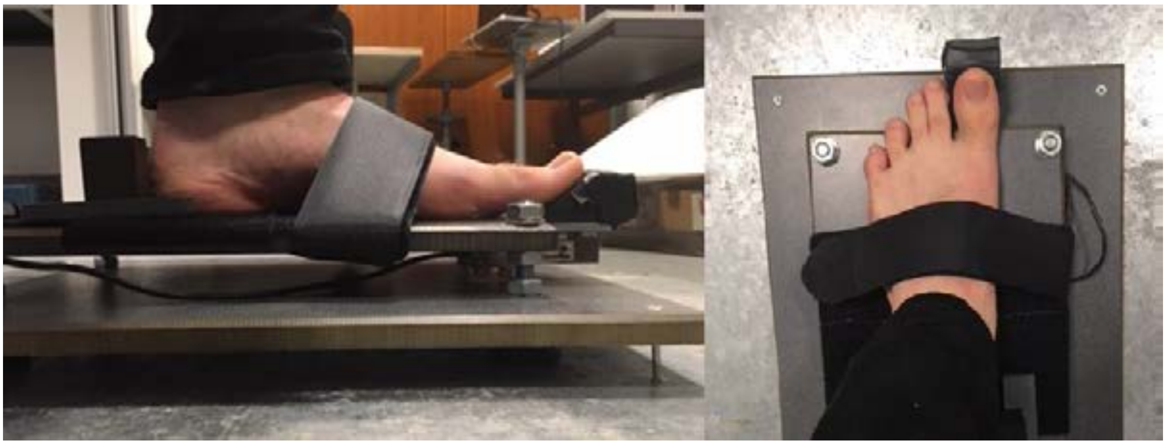

Figure 15 – A photo of the device used to assess great toe flexor strength.

The gTFM strength was assessed through a range of 1st metatarsal phalangeal joint (MPJ) and ankle joint angles. The combination of MPJ and ankle joint angles were 1) 0° dorsal flexion of the MPJ and 0° dorsal flexion of the ankle, 2) 35° dorsal flexion of the MPJ and 0° dorsal flexion of the ankle and 3) 0° dorsal flexion of the MPJ and 35° plantarflexion of the ankle. These positions were included as they covered a range of TFM force production outcomes as observed in a previous study investigating TFM strength (Goldmann and Bruggemann 2012). The procedure in this study involved the participant sitting upright with their arms across their chest. Their foot was strapped into the device (Figure 15) firmly but comfortably over the dorsal surface using Velcro® and a heel support was placed behind the calcaneus such that the plantar surface of their great toe was positioned onto a plate fixed to a strain gauge. The position of the device was then adjusted closer or further away from the participant to get the angle of their ankle to the correct position using a manual goniometer. By 0° dorsal flexion of the ankle we imply a neutral position of the ankle with the shank in an upright position with the knee above the ankle. To position the great toe in to 35° dorsal flexion a wedge was fixed on top of the strain gauge and the plantar surface of the great toe was placed onto the sloped surface. These angles were checked prior to each trial being completed. Each measurement consisted of them pushing down with their great toe using their gTFM without external influences. Participants were instructed to keep their heel on the base plate, maintain their leg and upper body position and keep their arms across their chest for the duration of the contraction and these were monitored by the investigator throughout each trial. Any trials where these conditions were not met were discarded. Prior to each measurement the participant was instructed to relax their foot and the signal caused by the passive structures acting on the strain gauge was reset to zero. Each participant completed 3 sub-maximal contractions on the device to familiarise themselves with the movement and the protocol and ensure they were comfortable before performing 3 maximal voluntary isometric contractions (MVIC). Each contraction lasted for 3 seconds verbalised through an audible cue by the investigator and a rest period of 1 minute was given in between contractions to prevent fatigue. Once the three MVIC were completed the participant removed their foot from the device and was given a 2 minute rest before repeating the procedure in another position. The participant repeated the procedure at each of the three angle combinations and with both feet in a randomised order (Set 1). The same procedure was then completed on another day separated by at least 24hours (Set 2).

The device itself consisted of a foot plate with adjustable heel support and Velcro straps to hold the foot in place (Figure 15). A strain gauge (NeuroLog NL62) was located at the end of the foot plate and could be adjusted in order to be placed beneath the plantar surface of the great toe. The strain gauge was connected to an amplifier and a data acquisition device (DAQ) with data collected through Mr Kick III software for Windows 7. Prior to data collection the device was calibrated with a set of calibration weights to determine the voltage-force equation. The maximum peak of the voltage signal during the 3 second contraction window was recorded for each successful trial and then converted into force (N) using the equation before being averaged for each set in each position for the dominant (Table 8) and non-dominant (Table 9) feet. . Forces were used as opposed to moments due to the difficulty in getting a true measurement of the moment arm from the point of force application to the strain gauge to the first MPJ joint space.

A paired samples t-test with Pearsons correlation coefficient was completed to determine how the results from Set 1 compared to Set 2. A mixed design repeated measures analysis of variance (ANOVA) with position as a within subject factor and foot dominance as the between subject factor was completed for both males and females to determine any differences between position and the effect of foot dominance. All statistical analysis was completed in SPSS V.22 for Windows (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$.

Results

The one leg standing test for leg dominance demonstrated that two thirds of both males (10/15) and females (10/15) chose their left foot to be their dominant side for stabilisation.

Males

Foot Dominance

There was no significant difference in gTFM strength between the dominant and non-dominant feet ($F(1,28) = 0.990$, $p=0.756$).

Male	Ankle 0°, 1MPJ 35° (N)	Ankle 35°, 1MPJ 0° (N)	Ankle 0°, 1MPJ 0° (N)
Set 1	147.03 (42.60)	80.66 (26.63)	129.75 (36.09)
Set 2	156.04 (34.56)	85.85 (23.09)	131.13 (34.19)
Female	Ankle 0°, 1MPJ 35° (N)	Ankle 35°, 1MPJ 0° (N)	Ankle 0°, 1MPJ 0° (N)
Set 1	82.29 (34.90)	51.43 (23.27)	63.36 (29.56)
Set 2	85.60 (37.06)	54.41 (22.55)	70.01 (32.61)

Table 8: A table to illustrate the force produced in each position and in each set by the great toe of the dominant foot for both males and females (Means (±S.D)).

Position

There was a significant difference between positions ($F(1.630,45.651) = 89.125$, $p<0.001$). Post-hoc comparisons illustrate that 1) significantly more force was produced in the Ankle 0°, 1MPJ 35° position compared to the other two positions, 2) significantly more force was produced in the Ankle 0°, 1MPJ 0° position than in the Ankle 35°, 1MPJ 0° position. There was no significant position*footdominance interaction observed ($F(1.630,45.651) = 0.348$, $p=0.664$).

Sets

The correlation analysis demonstrated that there was a high reproducibility of results between sets with correlation coefficient values of 0.948, 0.947 and 0.961 for the Ankle 0°, 1MPJ 35°, the Ankle 35°, 1MPJ 0° and the Ankle 0°, Toe 0° positions respectively (Figure 17).

Male	Ankle 0°, 1MPJ 35° (N)	Ankle 35°, 1MPJ 0° (N)	Ankle 0°, 1MPJ 0° (N)
Set 1	155.29 (53.94)	80.23 (29.86)	131.93 (57.39)

Set 2	159.34 (45.32)	84.16 (27.68)	144.42 (56.99)
Female	Ankle 0°, 1MPJ 35° (N)	Ankle 35°, 1MPJ 0° (N)	Ankle 0°, 1MPJ 0° (N)
Set 1	83.57 (25.57)	56.27 (21.72)	73.80 (24.74)
Set 2	85.61 (35.21)	57.07 (20.02)	73.30 (26.93)

Table 9: A table to illustrate the force produced in each position and in each set by the great toe of the non-dominant foot for both males and females (Means (±S.D)).

Females

Foot Dominance

There was no significant difference in gTFM strength between the dominant and non-dominant feet ($F(1,28) = 0.150$, $p=0.701$).

Position

There was a significant difference between positions ($F(2,56) = 53.424$, $p<0.001$). Post-hoc comparisons illustrate that 1) a significantly greater amount of force was produced in the Ankle 0°, 1MPJ 35° position compared to the other two positions, 2) significantly more force was produced in the Ankle 0°, 1MPJ 0° position than in the Ankle 35°, 1MPJ 0° position. There was no significant position*foot-dominance interaction observed ($F(2,56) = 0.596$, $p=0.554$).

Sets

The correlation analysis demonstrated that there was a high reproducibility of results between sets with correlation coefficient values of 0.947, 0.933 and 0.950 for the Ankle 0°, 1MPJ 35°, the Ankle 35°, 1MPJ 0° and the Ankle 0°, Toe 0° positions respectively (Figure 16).

Discussion

The results from this study demonstrate that the custom-built device is reliable in assessing the strength of the gTFM. It showed excellent reproducibility in measures between two sets of three

MVIC with Pearson's correlation coefficient values ranging between 0.933 and 0.961 for the different positions. This illustrates that with repeated measures, and following the same procedure, this device can be used to record the strength of the gTFM in future study. The results are also consistent with that from previous research investigating the force produced by the TFM as a function of joint angles (Goldmann and Bruggemann 2012). We observed the same relationship across joint angles with the highest forces being produced by the gTFM with the ankle in a neutral position (0°) and the great toe in dorsal flexion (35°), the lowest forces produced with the ankle in plantarflexion (35°) and the great toe in a neutral position (0°) and the position with the ankle in a neutral position (0°) and great toe in a neutral position (0°) resulted in an intermediate production of force. This further helps to support that this custom-built device is reliable at recording the forces produced by the gTFM.

The results of this study are explained by the force-length relationship of a muscle and the associated cross bridge theory. Muscles are able to function over a range of different lengths however they don't produce equal force across their whole range due to the amount of contractile element overlap and the limits of sarcomere length (Gordon, Huxley et al. 1966). The force-length relationship is therefore seen as an inverted U shape. Through altering joint angles the muscle length can be directly affected (Hawkins and Hull 1990). In this study, when the ankle was placed in 35° plantarflexion this had the effect of mechanically shortening the plantarflexors, including the flexor hallucis longus, placing them in a sub-optimal length and thus reduced force capacity. Conversely by placing the 1MPJ in a position of 35° dorsiflexion this would lengthen the flexor hallucis longus, counteracting the ankle joint position and situating it closer to its optimal length thus increasing the force capacity. It demonstrates how for muscles operating across multi-joints changing the joint angle configuration is inexplicably linked to the force-length relationship of those muscles.

This study can also provide support to the previous findings in the literature which explained that dorsal flexion at the metatarsophalangeal joints help to assist in the push off phase of locomotion to counteract the reduction in force producing capabilities when the ankle is plantar flexed (Goldmann and Bruggemann 2012). This could have relevance for introducing training protocols specifically aimed at increasing TFM strength to assist in this push-off phase of locomotion. It has been previously shown that increasing TFM strength can be beneficial for movement performance such as jumping and sprinting with significant improvements observed following a short TFM strength training program (Goldmann, Sanno et al. 2011, Hashimoto and Sakuraba 2014).

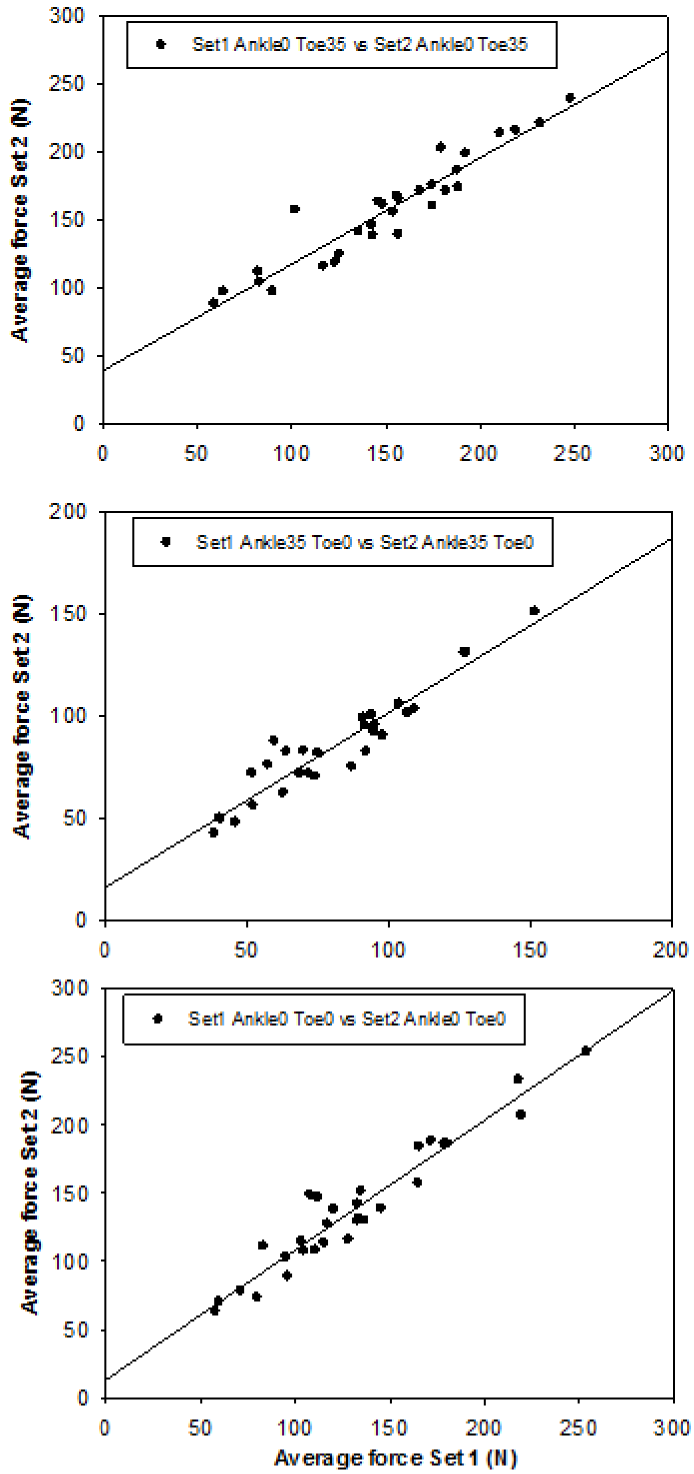

Figure 16: A correlation to illustrate the reproducibility of forces produced by the gTFM between two sets of 3 MVIC at each position for all female participants (r = 0.947, 0.933 and 0.950 respectively)

In these studies there were no improvements in walking witnessed however these were performed in young healthy and active individuals and thus any improvement is likely to be negligible. It is well documented that in older age there is often a decrease in step length and gait speed (Prince, Corriveau et al. 1997). It is also known that as we age there is a reduction in ankle plantar flexor power and thus the work is often shifted to more proximal structures such as the hips and knees putting extra strain on these areas (Franz and Kram 2013). At the push off phase in locomotion the ankle is placed in a mechanically inefficient position and as the results from this study, and previous studies, have shown, the TFM with the MPJ placed into dorsal flexion help to account for some of this deficit. It could therefore be worthwhile investigating if improving the strength and force producing capabilities preferentially in the TFM could lead to improvements in gait parameters in the older age population.

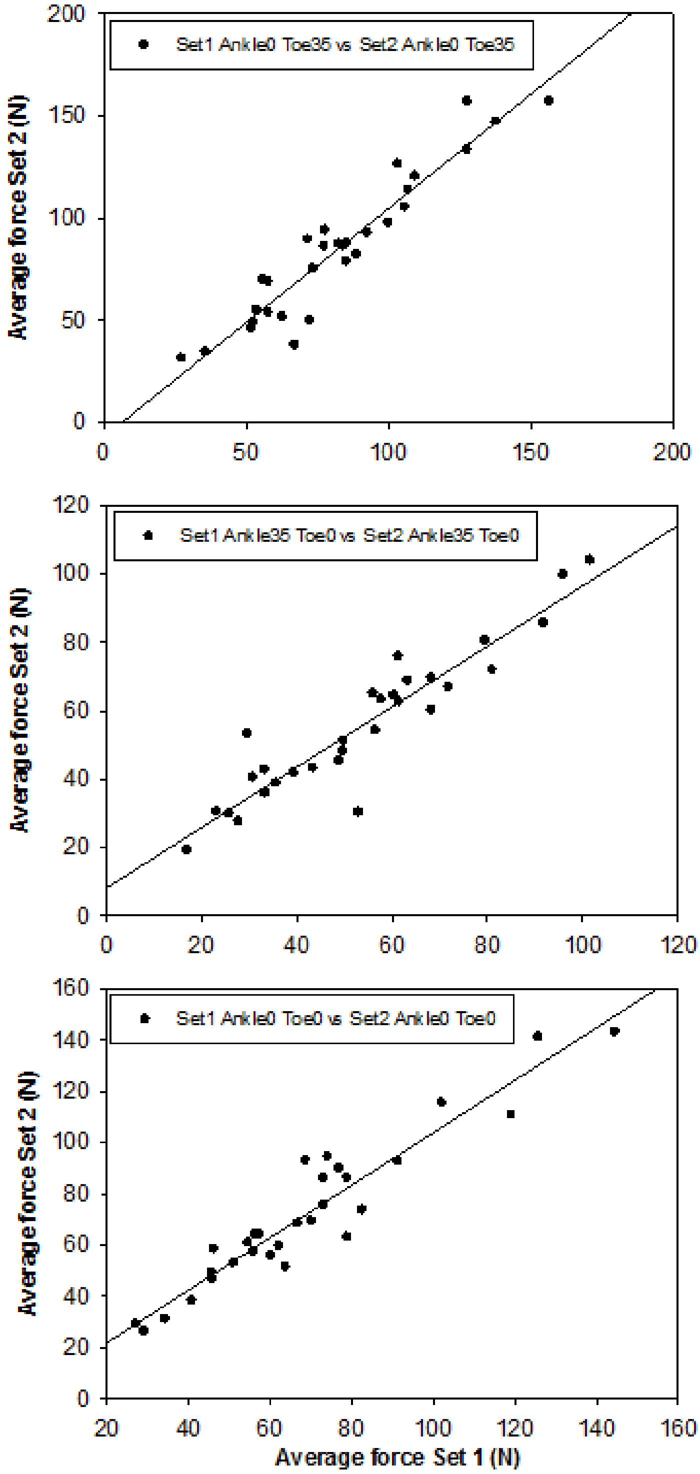

Figure 17: A correlation to illustrate the reproducibility of forces produced by the gTFM between two sets of 3 MVIC at each position for all male participants (r = 0.948, 0.947 and 0.961 respectively)

Aside from the impact on gait there is research indicating that the strength of the TFM play an important role in standing and reaching. Previous research has identified that increased TFM strength has a clear relationship with the anterior functional base of support and the corresponding reach or lean capacity of an individual (Endo, Ashton-Miller et al. 2002). This study also reported a 28.9% reduction in TFM strength in a group of older individuals compared to their younger counterparts. This reduction in TFM strength is suggested to negatively contribute to elderly individuals' ability to perform certain reaching and leaning tasks whilst maintaining a safe degree of stability. It seems reasonable to suggest that working to improve TFM strength could have benefits for improving the elderly's ability to perform certain activities of daily living and reducing the risk of instability and falls. With this in mind, from a methodological point of view, prior research has indicated that commonly used static balance tasks such as quiet stance COP measures on a force plate are unrelated to functional reaching and more dynamic balance tasks and thus do not give an overall indication of postural control (Riemann and Schmitz 2012). As TFM strength is specifically important for tasks such as reaching and leaning, it is therefore essential for future studies looking to investigate the effect of TFM strength on balance to include functional balance tests as part of their testing protocol.

A final result of this study which requires highlighting is there was no difference in gTFM strength between the dominant and non-dominant feet in this population. In this study, in both males and females, two thirds of the participants chose to stand on their left leg for the one legged stance task of foot dominance. This is consistent with previous findings which showed a 60% tendency to choose the left side for one legged stance (Velotta, Weyer et al. 2011). Although there is a clear preference for one side to be dominant there doesn't appear to be any corresponding differences in gTFM strength in this population which reflects that dominance. It is unknown whether in other populations where limb dominance is more pronounced, and differences in postural balance ability between the dominant and non-dominant sides are present e.g. soccer players (Barone, Macaluso et al. 2010), if there would then be a corresponding disparity in TFM strength.

Conclusion

In conclusion this study has demonstrated that the customised device is reliable at measuring gTFM strength and can thus be used in future study. It has also supported previous findings that dorsiflexion at the MPJ counteracts some of the mechanical inefficiency and loss of force experienced with the ankle in plantarflexion. This is particularly relevant at the push off phase of the gait cycle and illustrates the importance of maintaining TFM strength for efficient locomotion. The importance of TFM strength in functional reaching tasks was also discussed and the potential benefit of improving TFM strength particularly in the elderly population was introduced.

Chapter 6

Does increased intrinsic foot strength lead to improvements in balance and gait in older adults?

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Abstract

Ageing often results in declines in sensory and motor function contributing to an increased risk of postural instability and falls (NICE 2013). The feet serve as our connection to the ground and the foot muscles play a role in balance control (Endo, Ashton-Miller et al. 2002, Spink, Fotoohabadi et al. 2011) and gait (Peter, Hegyi et al. 2015) with toe weakness being reported as a fall predictor (Spink, Fotoohabadi et al. 2011, Mickle, Angin et al. 2016). The aim of this study was to investigate if increasing intrinsic foot muscle strength via a 6-week home-based seated foot exercise program improves foot posture, balance and gait of older adults. Nineteen healthy older adults (4 male; 70.5±3.3 years) participated in a 6-week home-based daily foot strength training program. Participants were assessed on hallux plantar flexor strength, foot mobility, postural sway, dynamic balance and gait performance at baseline and following the 6-week intervention. The training consisted of 4 exercises comprising toe flexion, toe extension, toe abduction and a toe grasping task. Paired samples t-tests were used to assess changes from baseline to follow-up. This exercise program was shown to effectively increase foot strength (maximal hallux plantar flexor strength: baseline: 42.4N±16.7, follow-up: 55.8N±18.3). Following the 6 weeks of foot strength training, there was a greater functional base of support (functional reach test performance: baseline: 29.0cm±5.4, follow-up: 31.9cm±4.3) as well as, during quiet bipedal stance, a reduction in medial lateral (ML) sway velocity (centre of pressure: baseline: 23mm/sec±7, follow-up:20mm/sec±7), ML sway displacement (centre of pressure: baseline: 11mm±2, follow-up: 9mm±3) and total sway path length (centre of pressure: 899mm±280,

follow-up: 816mm±255). There were no significant changes in gait parameters. Home-based seated foot exercises are effective at promoting foot muscle strength and can be employed to target foot weakness and improve balance in older adults.

Introduction

Falling is a consistent problem in older age, with approximately 1 in 3 adults over 65 years old suffering at least one fall per year (NICE 2013). This susceptibility to falls in older age is multi dimensional and a combination of factors far beyond the scope of this article however, one risk factor to be identified in recent research and consequently providing the focus of this study is reduced foot strength; in particular the toe flexors (Menz, Morris et al. 2005, Mickle, Munro et al. 2009, Spink, Fotoohabadi et al. 2011, Yong-Wook, Oh-Yun et al. 2011). Spink et al (Spink, Fotoohabadi et al. 2011) observed in a large cohort of older adults hallux plantar flexion strength and ankle inversion-eversion range of motion to be the most consistent independent predictors of balance ability. Similarly in a separate study following a large group of older adults over a 12 month period to establish fall incidence, fallers displayed significantly reduced hallux and lesser toe strength compared to non-fallers (Mickle, Munro et al. 2009). The authors reported that based on the strength of the hallux alongside the presence of lesser toe deformities they could predict 64% of fallers whilst a 1% bodyweight increase in hallux strength decreased fall risk by approximately 7% (Mickle, Munro et al. 2009).

Foot problems are also a common problem in older ages which have also been reported as an independent risk factor for falling (Menz and Lord 2001, Mickle, Munro et al. 2009). Hallux valgus, one of the most common foot problems in this population reportedly present in up to 33-44% of women and 14-25% of men (Nix, Smith et al. 2010), has been reported to have a detrimental effect on gait stability through modified plantar loading (Galica, Hagedorn et al. 2013), reduced propulsive power and arrhythmic acceleration patterns (Menz and Lord 2005). One of the reported contributing factors to foot problems such as hallux valgus could be reduced intrinsic foot muscle strength due to their observed role in maintaining foot posture (Fiolkowski, Brunt et al. 2003, Headlee, Leonard et al. 2008, Kelly, Cresswell et al. 2014). A significant decrease in Abductor Hallucis (ABH) size was witnessed in older adults suffering from hallux valgus (Aiyer, Stewart et al. 2015) whereas in a separate older adult population without toe deformities no decrease in ABH size was observed suggesting a potentially protective effect against hallux valgus by maintaining ABH strength (Mickle, Angin et al. 2016).

The toes, and associated muscles, have been shown to play a prominent role during walking (Peter, Hegyi et al. 2015, Hughes, Clark et al. 1990) and in maintaining stability during tasks whereby the centre of mass approaches the edges of the functional base of support such as leaning and reaching (Endo, Ashton-Miller et al. 2002, Yong-Wook, Oh-Yun et al. 2011). During the push off phase of walking, the ankle undergoes plantar flexion and the heel lifts of the ground. The toe flexors maintain high forces keeping the toes in contact with the ground serving to increase the effective leg length and stabilising the body (Whittle 2002). At this terminal stance phase an estimated 24% of body weight is placed on the hallux (Jacob 2001) highlighting the necessity for sufficient toe flexor strength. Similarly, the anterior limit of stability has been shown to be significantly correlated to hallux flexor strength leading to a decline in the functional base of support (Endo, Ashton-Miller et al. 2002). Hallux flexor strength is known to decrease considerably with age (Mickle, Angin et al. 2016) resulting in the potential increased risk of falling during activities of daily living involving leaning or reaching (Endo, Ashton-Miller et al. 2002).

There have been previous intervention studies targeting increasing strength in the lower extremity and positive effects of these interventions on fall risk and balance ability have been observed. Recently a supervised 12 week progressive resistance exercise program was implemented in a sample of older adults and compared to a home based exercise group and a control group to determine the effectiveness of each intervention at increasing hallux and lesser toe flexor strength (Mickle, Caputi et al. 2016). It was observed that the progressive exercise program resulted in up to a 36% increase in toe flexor strength whilst the home based program appeared ineffective and showed no change (Mickle, Caputi et al. 2016). The authors indicate that the success of an exercise program at increasing toe flexor strength is dependent on having a progressive element and resistance bands are particularly convenient to encourage this.

Whilst this study by Mickle et al. (Mickle, Caputi et al. 2016) demonstrates the benefits of a supervised exercise program it is clear that if similar results could be achieved in a home based setting this would be potentially more advantageous to a wider population particularly for those with travel or time restrictions. This study therefore attempts to design a simple but effective home based seated exercise program, encompassing a progressive element, to increase the foot strength of older adults. It also aims to determine if increased foot strength improves foot posture, balance and gait. We hypothesised that an increase in foot strength would be observed, corresponding with: 1) an improvement in foot posture measured by a reduction in the foot mobility magnitude; 2) an improvement in balance indicated by an increase in functional reach performance and a decrease in postural sway parameters; and 3) an improvement in gait performance indicated by a faster walking speed and increased ankle plantar flexion at toe off.

Methods

20 older adults (5 male; 70.7±3.4 years, Body Mass Index (BMI) 24.13±3.56 kg/m^2) volunteered to participate. All were healthy, had no known gait disorders or suffered from diabetes or disorders that affect foot sensation and none had recently participated in any other targeted foot strength or balance training. Prior to participation all participants were provided with an information sheet about the study and signed an informed consent and a health screening questionnaire as per the University Science, Technology, Engineering and Mathematics Ethical Review Committee requirements (ERN_09-528AP14, approved 22/04/2016).

All participants visited the laboratory on 2 occasions, baseline and after a 6-week training program. During each visit participants underwent tests of static and dynamic balance, gait performance, foot mobility and hallux plantar flexor strength completed by the same experimenter. At the start of each testing visit, all participants' height and weight were measured whilst barefoot and lightly clothed.

A postural sway task was used to assess static balance. This test required standing on a force plate with feet in a parallel stance such that the medial edges were touching and the heels and toes were aligned. All participants were instructed to stand upright, place their hands on their hips and look directly forwards before closing their eyes. They were instructed to remain standing as still as possible with their eyes closed for the duration of the trial. Each trial commenced once the eyes were closed and lasted for 35seconds. This was repeated 3 times so that an average could be taken, as has been deemed a reliable procedure when assessing body sway (Sarabon, Rosker et al. 2010). The force plate (Kistler Type 9281C, dimensions 600x400mm) recorded at 1000Hz, and the centre of pressure (COP) trajectory in the medial-lateral (ML) and anterior-posterior (AP) directions were calculated from the force by in-built algorithms in the Bioware software used for data collection. This was low pass filtered using a zero-lag fourth order Butterworth filter at 8Hz. The resultant displacement (Res) was then calculated using the following equation $Res = \sqrt{(ML^2) + AP^2}$ to represent the overall COP path as an indicator of the centre of mass in relation to the base of support. Average displacement, range and velocity for each direction across each 35second trial was then calculated and an average of the 3 trials was taken. Sway path length was calculated by calculating the difference between each displacement sample from the Res trajectory (i.e. the resultant displacement from one sample to the next) and summing to calculate the total length travelled from the initial starting position. Sway area was approximated by using principal component analysis (PCA) to plot a prediction ellipse around the COPx and COPy data and calculating the prediction ellipse area with a probability value set at 0.9. The principal component analysis involved calculating the eigenvalues and eigenvectors from the covariant matrix using all the ML and AP data points and the mean values. The data points were orientated such that they were centred about the origin and the maximum and minimum values were found. The prediction ellipse was then fitted using the eigenvalues specifying the main axes of the ellipse and the area of the fitted ellipse calculated as an estimation of the sway area. All data analysis was completed using Matlab (MATLAB, The MathWorks, Natick, MA, USA).

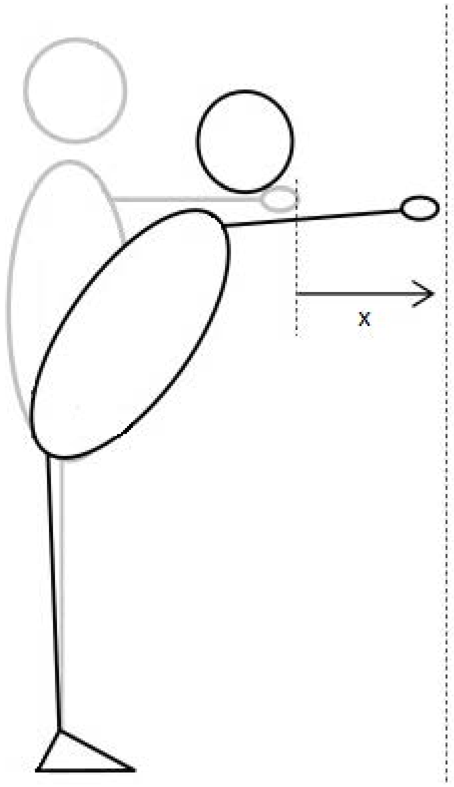


Figure 18 Functional Reach Test (FRT). X signifies the computed distance between the head of the 3rd metacarpal at the start upright position and the furthest point the participant could reach along the wall. Modified from (Kage, Okuda et al. 2009).

The functional reach test was used to assess dynamic balance and forward limit of stability as previously established and validated (Duncan, Weiner et al. 1990, Martins, de Menezes et al. 2012).. The task involves standing with the feet shoulder width apart parallel to a wall with a measuring stick affixed horizontally at the height of their shoulder. Whilst standing in an upright position and with no torso rotation, the arm closest to the wall is raised to 90° of shoulder flexion with the hand in a closed fist. This position was deemed the starting point, and the position of their 3rd metacarpal head along the measuring stick is recorded. The participant then reaches as far forward along the measuring stick as possible without moving their feet or losing balance. The furthest point they could reach was again recorded and the distance between points was calculated. Each participant was demonstrated the task by the experimenter and given a practice trial before completing 3 trials on each side of the body and an average taken.

Gait performance was assessed using repeated 10m walking trials. These were completed in two footwear conditions: barefoot and in a pair of standardised shoes (Style Code: 10001, Hobos

Womens, Style Code: 50109, Hobos Mens). Participants completed 3 trials at their preferred walking speed and 3 trials at their fastest walking speed in both footwear conditions. These were administered in a counter-balanced order across participants but in the same order at baseline and follow-up to reduce the time spent changing footwear and the possibility of error by minimising the number of marker placements. Participants started their 10m walking trial at a pre-determined mark which allowed for at least 3 steps prior to the recording field of view and stopped at another mark positioned outside of the recording field of view to allow for acceleration and deceleration effects. To record gait kinematics, a Vicon MX system comprised of 13 cameras recording at a sampling rate of 100Hz was used which had a recording field of view covering the middle 6m of the 10m walking lane. Reflective markers were placed on the participant's lower body using adhesive tape in the positions according to the Vicon Plug-in gait lower body template. Calibration of the system was completed prior to data collection such that there was a residual error of less than 1mm. Post-processing of the data was completed using custom-written scripts in Matlab (MATLAB, The MathWorks, Natick, MA, USA). Kinematic data was interpolated to fill small gaps in trajectories and then low pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12Hz. Gait speed, step length, ankle angle at toe off and heel velocity at toe off were determined.

Foot mobility was assessed using the previously deemed reliable measure of foot mobility magnitude (FMM). An outline of the method used to correctly calculate this can be found elsewhere (McPoil, Vicenzino et al. 2009) but effectively involves measuring foot width and dorsum height at 50% of the total foot length under both weight bearing and non-weight bearing conditions to determine the change in foot shape. Plantar flexor strength of the hallux was assessed using a custom built device previously tested for reliability and demonstrating correlation coefficient values between visits of $r = 0.95$. The participant's foot was placed into a position of 35° ankle plantar flexion, and their hallux in a position of 35° dorsiflexion and their foot secured with Velcro® straps. This position was chosen to simulate the role of the hallux during the final point of toe off during walking and to limit the external influence of larger ankle plantar flexors (Soysa, Hiller et al. 2012) with previous research highlighting how hallux dorsiflexion counteracts the reduction in force producing capabilities when the ankle is plantar flexed (Goldmann and Bruggemann 2012). The participant was instructed to maintain this foot and leg position, sit upright on the chair with their arms across their chest. The task involved pushing down with their hallux onto the wedge affixed on top of a strain-gauge (NeuroLog NL62). The raw data was amplified and digitized using an analog-to-digital card (National Instruments BNC-2090A, National Instruments USB-6251), and the data sampled by Mr Kick III software for Windows 7. The strain gauge was calibrated using a set of calibration weights prior to data collection and a calibration equation was determined to convert voltage (V) into force (N). Three sub-maximal practice trials were given to each participant to ensure they were comfortable with the task and feedback provided by the experimenter if required. A set of three recorded maximum voluntary isometric contractions (MVIC) lasting 3-5 seconds followed with a recovery period of 1 minute in between repetitions. The same procedure was then completed with the other foot. The maximum force as well as the impulse during the first 3 seconds following the initial rise in force was computed. These testing protocols were completed at baseline and then repeated at the 6-week follow-up session.

During the 6 week period between baseline and follow-up tests, participants were provided with a set of foot strength training exercises to be completed daily. They were provided with a training diary to document their training and which outlined the exercises to be completed (Table 10). The protocol consisted of 4 simple exercises that were to be completed at home at a time of most convenience and in a seated position. They were designed to enable the participant to easily fit them into their daily routine to ensure adherence. The aims of the exercises were to increase the strength of the intrinsic foot muscles and improve foot/toe mobility and consisted of maximal plantar flexion, dorsiflexion and abduction of the toes and a toe grasping task. The first exercise involved maximally plantar-flexing their hallux into the ground whilst simultaneously dorsi-flexing their 4 lesser toes off the ground. Assessed using MRI, completing this exercise has been shown to result in a 15.1% increase in flexor digitorum brevis (FDB) activation, 16.5% increase in abductor hallucis (ABH), 22.5% increase in abductor digit minimi (ADM), 20.2% increase in flexor hallucis brevis (FHB) and 25.4% increase in quadratus plantae (QP) (Gooding, Feger et al. 2016). The second exercise involved maximally plantar-flexing their 4 lesser toes into the ground whilst simultaneously dorsi-flexing their hallux off the ground. This exercise has been shown to increase activation by 18.1% in FDB, 16.9% in ABH, 14.1% in ADM, 13.1% in FHB and 13.9% in QP (Gooding, Feger et al. 2016). These two exercises involved holding this position with maximal effort starting at 5seconds and increasing by 3 seconds if successful for 3 days in a row up to a maximum of 15 seconds and repeat 3 times on each foot. The third exercise involved spreading toes through maximum abduction, holding this position for 3 seconds and then relaxing. This exercise has been previously shown to increase activation by 27.0% in FDB, 18.9% in ABH, 35.2% in ADM, 29.5% in FHB and 25.4% in QP(Gooding, Feger et al. 2016).

Exercises Description	Intensity	Illustration
Maximal Hallux Extension + Maximal Lesser Toe Flexion Raise 1st toe off the ground whilst pushing four outer toes down as hard as you can into the ground. Keep heel and forefoot in contact with the ground. Hold this position maintaining force as best you can. Rest for 10seconds in between repetitions.	Start 3 x 5secs on each foot. Progression through increasing length of time (+3secs) up to maximum of 3 x 15 secs on each foot.	
Maximal Hallux Flexion + Maximal Lesser Toe Extension Raise 4 outer toes off the ground whilst pushing 1st toe down as hard as you can into the ground. Keep heel and forefoot	Start 3 x 5secs on each foot. Progression through increasing length of time (+3secs) up to	

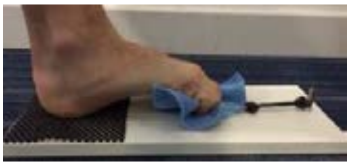
in contact with the ground. Hold this position maintaining force as best you can. Rest for 10seconds in between repetitions.	maximum of 3 x 15 secs on each foot.	
Maximal Toe Abduction Spread your toes apart so you can see space between each of your toes. Try to also move your 1st toe away from your 2nd toe. Hold this position for 3-5seconds, release then repeat.	Start 3 sets of 10 repetitions on each foot. Progression through increasing number of repetitions (+1 rep) up to maximum of 3 sets of 15 repetitions on each foot.	
Towel Grab Position your foot on the board such that your toes are in the centre of the cloth. Keeping your heel and foot stable on the black non-slip mat, grab the cloth with your toes and pull the cloth back towards you and hold for 3-5seconds, release and then repeat.	Start 3 sets of 10 repetitions on each foot. Progressions through increasing number of repetitions (+1rep) up to maximum of 3 sets of 15 repetitions on each foot then switch to stiffer resistance band.	 

Table 10 – A example of the exercises and progression comprising the foot strength training program to be completed by participants daily.

This was to be repeated for a maximum of 3 sets of 15 repetitions on each foot progressing up (+1 rep every 3 days) from a starting point of 3 sets of 10 repetitions. The final exercise used a custom-made toe flexion towel grab board. The board consisted of a base plate with a non-slip surface for the foot and a bar placed at one end of the board around which a resistance band was affixed. The resistance band was then attached to a small towel. The task involved placing the towel under the forefoot and then, without moving the heel backwards, curling the foot, raising the arch actively whilst grabbing the towel with the toes and pulling it back towards the body as far as possible creating resistance in the band. Participants had to hold this active position against the resistance for 3 seconds before slowly releasing the tension in the band back to a foot flat relaxed position. This was then repeated for 3 sets of 10 repetitions at the start to up to a maximum of 3 sets of 15 repetitions on each foot with the recovery for each foot being the time taken to complete a set on the other foot. In order to add a further progressive element to the exercises participants were provided with two bands with varying resistance (red and blue or blue and black depending on baseline strength) which they were instructed to move to the stiffer band once they could successfully complete 3 sets of 15 repetitions for 3 consecutive days. It took approximately 30 minutes each day for participants to complete these 4 exercises on both feet.

Paired samples t-tests were used to compare baseline to follow-up values for each variable with significance levels set to $p<0.05$. All statistical analyses were completed in SPSS v.22 for Windows (IBM Corporation, Somers, NY).

Results

One participant dropped out of the study midway through the 6-week intervention due to the towel grab exercise exacerbating their osteoathritic hallux pain. The results reported are from the 19 remaining older adults who completed the study (4male; 70.5±3.3years, BMI 23.98±3.60 kg/m^2).

Nine of the 19 participants completed the full training session on every day of the 6-week intervention period, with the remaining 10 participants missing on average 2.9±1.7 sessions out of the 42 scheduled sessions across the 6 weeks. The participants therefore reported a high rate of adherence with 86-100% of the sessions completed. All participants reached the maximal number of sets and repetitions across all exercises within the 6 week training program with 10 of the 19 participants also being able to progress to the higher intensity resistance band.

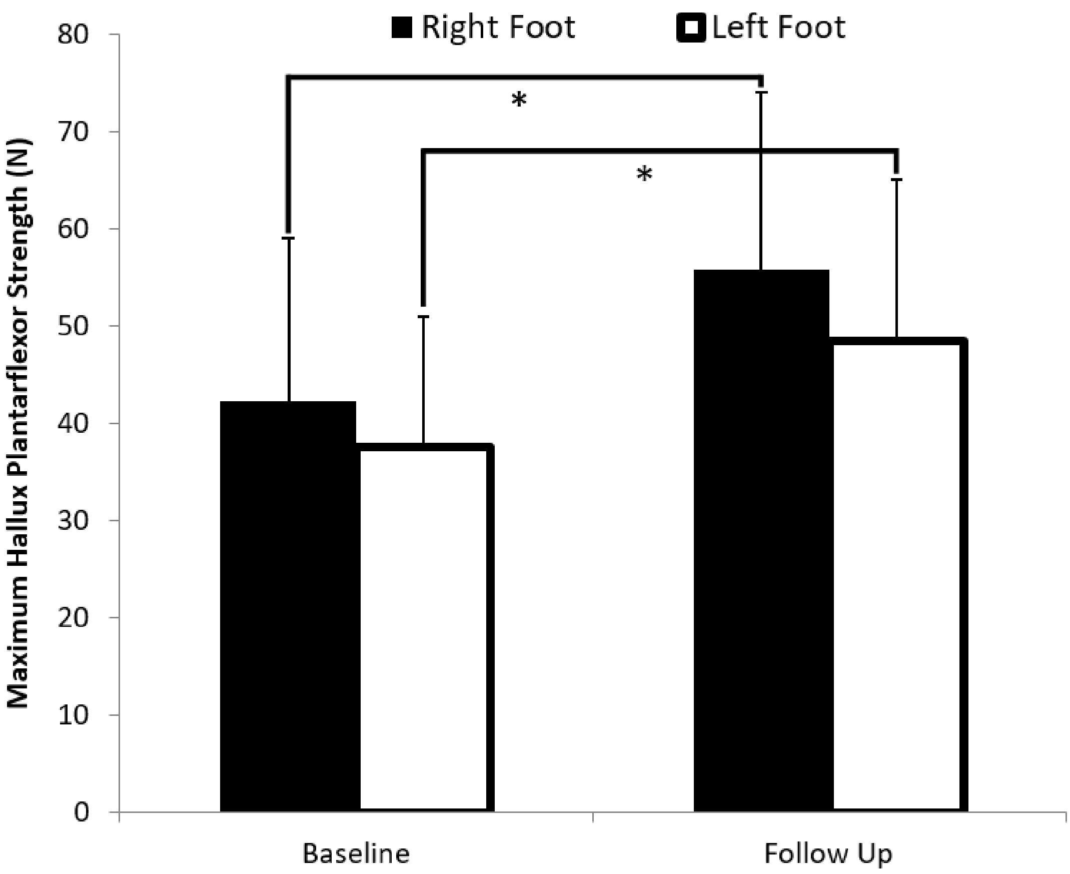


Figure 19 - The change in mean hallux plantar flexor maximal strength following a 6-week foot strength training intervention for the right and left feet. * indicates a significant difference from baseline to follow-up (p<0.05).

Hallux Plantar Flexor Strength

There was a significant increase in plantar flexor strength of both the right hallux (baseline: 42.4N±16.7, follow-up: 55.8N±18.3; $t(18)=4.24, p<0.001$) and left hallux (baseline: 37.6N±13.5, follow-up: 48.5N±16.5; $t(18)=5.06, p=0.001$) following the 6 weeks of foot strength training (Figure 19).

Right Foot	Baseline (mean±SD) (cm)	Follow Up (mean±SD) (cm)
Total Foot Length (TFL)	24.8±1.7	24.8±1.7
Dorsal Arch Height WB (50% TFL)	6.2±0.7	6.2±0.6
Midfoot Width WB (50% TFL)	8.2±0.6	8.2±0.6

Diff. Dorsal Arch Height (NWB-WB)	1.1±0.3	1.0±0.2
Diff Midfoot Width (WB-NWB)	0.8±0.3	0.7±0.2
Foot Mobility Magnitude	1.4±0.3	1.2±0.3*
Left Foot	Baseline (mean±SD) (cm)	Follow Up (mean±SD) (cm)
Total Foot Length (TFL)	24.8±1.8	24.8±1.8
Dorsal Arch Height WB (50% TFL)	6.2±0.8	6.3±0.7
Midfoot Width WB (50% TFL)	8.4±0.6	8.4±0.6
Diff. Dorsal Arch Height (NWB-WB)	1.3±0.4	1.2±0.3
Diff. Midfoot Width (WB-NWB)	0.7±0.3	0.7±0.3
Foot Mobility Magnitude (FMM)	1.5±0.4	1.4±0.3

Table 11 - Displaying the foot posture and foot mobility measures taken at baseline and follow up (baseline + 6weeks foot strength training). TFL = total foot length, WB = weight bearing, NWB = non-weight bearing, FMM = foot mobility magnitude, SD = standard deviation, * = significant difference from baseline to follow up (p<0.05).

Foot Posture and Mobility

As displayed in Table 11 there was a statistically significant decrease in the FMM score in the right foot (t(18)=2.13,p=0.047) but not significantly in the left foot (t(18)=1.31,p=0.208). It must be noted however that although statistically significant the change is smaller than the minimal detectable difference reported for this measure (McPoil, Vicenzino et al. 2009) and as such may not be clinically meaningful.

Dynamic Balance

There was a significant improvement in functional reach test performance on the right side (baseline: 29.0cm±5.4, follow-up: 31.9cm±4.3; t(18)=3.09,p=0.006), but not significantly on the left side (baseline: 30.0cm±4.4, follow-up: 31.1cm±4.2; t(18)=1.19,p=0.249) following the 6 weeks of training.

Static Balance

There was a significant improvement in average medial lateral (ML) sway displacement (Figure 20)(baseline: 11mm±2, follow-up: 9mm±3; $t(18)=3.18, p=0.005$), ML sway range (baseline: 45mm±11, follow-up: 41mm±10; $t(18)=2.41, p=0.027$), ML sway velocity (Figure 21) (baseline:23mm/sec±7, follow-up:20mm/sec±7; $t(18)=3.52$, $p=0.002$) and total sway path length (Figure 22) (baseline: 899mm±280, follow-up: 816mm±255; $t(18)=2.43, p=0.026$) at the 6-week follow-up session. Conversely there were no significant changes in the average anterior-posterior (AP) sway displacement (baseline: 10mm±4, follow-up:10mm±3; $t(18)=0.03, p=0.978$), AP sway range (baseline: 39mm±12, follow-up: 37mm±8; $t(18)=0.75, p=0.465$), AP sway velocity (baseline: 17mm/sec±6, follow-up: 16mm/sec±5; $t(18)=1.25, p=0.226$) or sway area (baseline: 753mm^2±371, follow-up: 679mm^2±310; $t(18)=1.41, p=0.176$).

Gait

There were no significant changes from baseline to follow-up in any of the gait parameters assessed when barefoot (BF) or in the standardised shoes (SH). These included gait speed (BF: baseline: 1.39m/sec±0.22, follow up: 1.39m/sec±0.21; SH: baseline: 1.47m/sec±0.18, follow up: 1.49m/sec±0.19), step length (BF: baseline:66cm±7, follow up: 67cm±7; SH: baseline: 72cm±6, follow up: 73cm±7) and ankle angle at toe off (BF: baseline: 105.4°±8.1, follow up:106.0° ±6.9; SH: baseline: 102.2°±8.0, follow up: 103.4°±7.7).

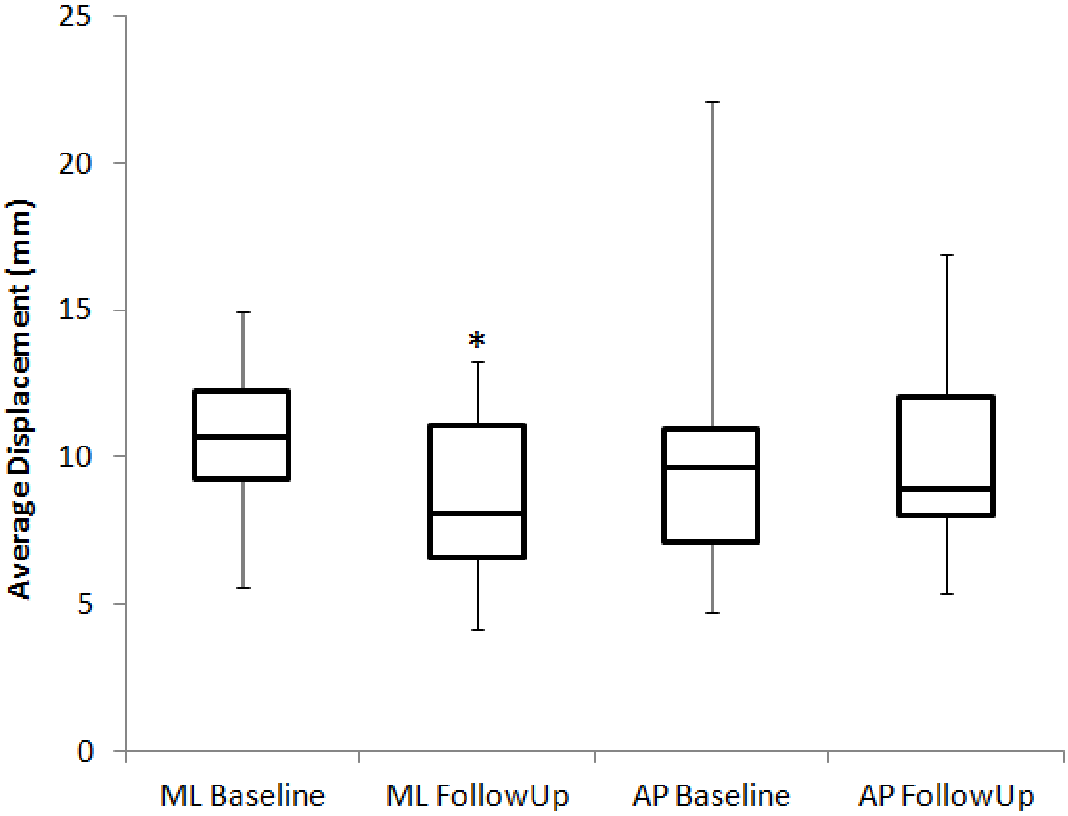


Figure 20 - The change in average displacement in the Medial Lateral (ML) and Anterior Posterior (AP) directions during a static postural sway task before and after a 6-week foot strength training intervention. * indicates a significant decrease from baseline to follow up ($p<0.05$).

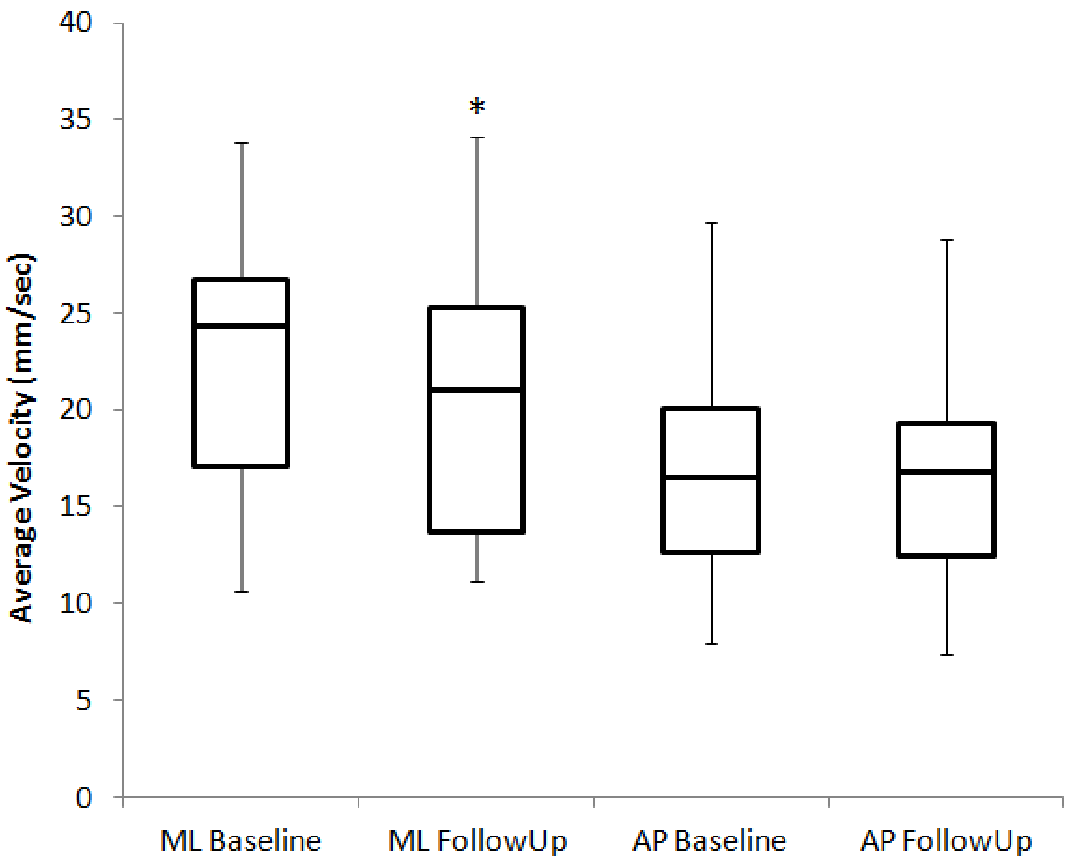


Figure 21 - The change in average Medial Lateral (ML) and Anterior Posterior (AP) sway velocity during a static postural sway task before and after a 6-week foot strength training intervention. * indicates a significant decrease from baseline to follow up (p<0.05).

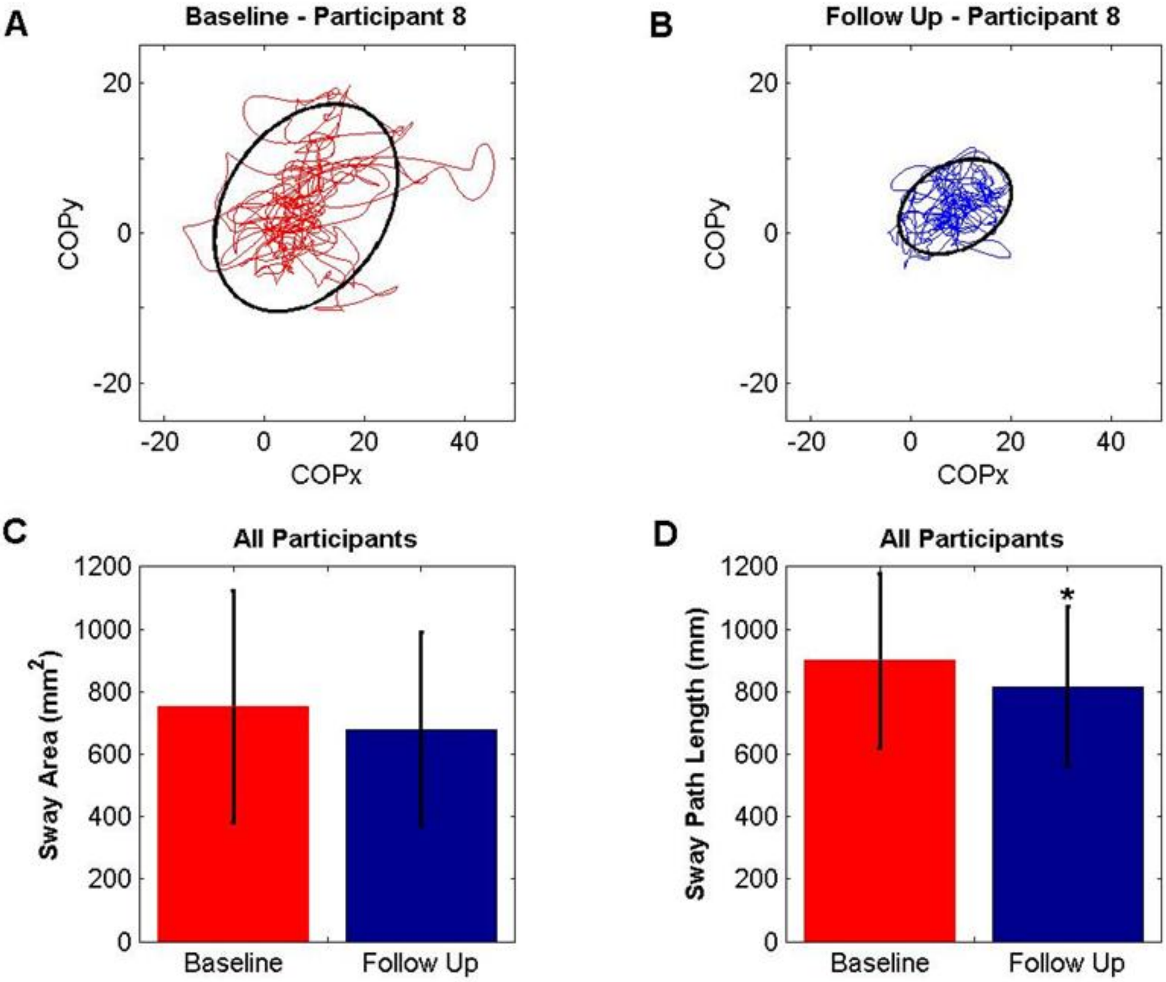


Figure 22 - Top: Stabilograms to display the centre of pressure (COP) trajectory for the last of 3 trials at baseline (A) and at the 6-week follow up (B) for one participant. Bottom: The bar graphs illustrate the mean sway area (C) and mean sway path length (D) at baseline and follow up for all participants with error bars indicating the standard deviation. *indicates a significant difference from baseline to follow up (p<0.05).

Discussion

This study aimed to determine if the strength of intrinsic foot muscles could be increased in older adults who followed 6 weeks of home-based foot strength training and if so, whether this contributed to improvements in foot posture, balance and gait. The results from this study showed that older adults did improve their foot strength, assessed via hallux plantar flexor MVIC, following 6 weeks of daily foot strength training at home (Figure 19). We can therefore accept our first hypothesis, whereby this strength training protocol is deemed successful in promoting foot strength increases in this older adult population. Furthermore as hypothesised, the foot strength increases coincided with an improvement in dynamic balance with a greater forward limit of stability, as well as reduced postural ML sway (Figure 20), ML sway velocity (Figure 21) and total sway path length (Figure 22) during quiet stance. Conversely, the final hypothesis was rejected as no changes in gait measures were observed in the study.

The improvements in foot strength observed as a result of the 6-week training program are promising given that this was a home- based protocol with the exercises able to be completed in a seated position at a time most convenient for participants. The protocol was designed in this way to ensure that it was easy and accessible for all, and would be suitable for the majority of the older adult population of which some may be restricted by mobility or travel constraints. The high rate of adherence (86-100% of total sessions completed) to the training program demonstrates that these exercises can be used in future studies targeting improved foot strength in the older adult population.

Although deemed statistically significant there were no clinically meaningful changes in foot posture changes witnessed in this study; the reduction in FMM score from baseline to follow-up in the right foot was not deemed greater than the minimal detectable change (0.2) reported for this measure (McPoil, Vicenzino et al. 2009). This therefore suggests that it is likely that foot posture did not change considerably as a result of this foot strength training protocol. It is well known that the longitudinal arch, and thus the vertical structure of the foot, is supported by the intrinsic foot muscles under tasks of greater postural demand (Fiolkowski, Brunt et al. 2003, Kelly, Cresswell et al. 2014) and therefore improvements in foot posture as a result of the increased foot strength may have been observed in a more challenging postural task than the bipedal stance employed in the FMM assessment.

Weakness in these supportive muscles within the foot is also associated with foot problems (Aiyer, Stewart et al. 2015) and foot pain (Latey, Burns et al. 2017), which are especially prevalent within the older adult population (Nix, Smith et al. 2010, Menz, Dufour et al. 2013). These problems can have a severe impact on daily living, with a recent review highlighting that 20% of older adults state foot pain to be the primary cause of staying indoors (Menz 2016). It is therefore paramount to ensure that interventions such as this one or others (Mickle, Caputi et al. 2016) are put in place in order to maintain strong foot function and health to preserve mobility and reduce the risk of developing disabling foot pain and its associated complications in later life.

Foot muscle strength has been shown to decline with age, and this coincides with a reduction in the functional base of support (Endo, Ashton-Miller et al. 2002). This study has further supported this relationship, and has shown how this can be reversed with a targeted foot strength training program. A significant improvement in functional reach test performance corresponding to improvements in foot strength were observed after just 6 weeks of foot strength training, indicating an increase in the forward limit of stability and base of support. Offsetting this age-related decline in muscle strength with interventions such as this one could serve to help preserve the functional base of support and reduce balance deficits.

Improvements in postural balance were also observed following the increase in intrinsic foot muscle strength. The decreases witnessed in ML sway displacement, ML range, ML velocity and sway path length are associated with balance improvements and a reduction in fall risk (Melzer, Benjuya et al. 2004). These data are in accordance with a previous study suggesting improved toe/foot strength is related to improved postural balance (Kobayashi, Hosoda et al. 1999). The results of this study showed no significant change in AP sway but this could primarily be due to the stance position. The feet together parallel stance largely evokes imbalance in the ML direction rather than in the AP direction as can be seen by the sway path trajectory and the orientation of the ellipse in Figure 22. Consequently differences in sway from baseline to follow up were most likely to be observable in the ML direction, where the instability is present, compared to the AP as the results appear to indicate. Improving ML stability in older adults is of major relevance due to the six fold increase of experiencing a hip fracture with a sideways fall (Greenspan, Myers et al. 1994) and thus these findings are particularly promising.

Alongside promoting strength increases, a previous study (Gooding, Feger et al. 2016) has demonstrated that the exercises used in this intervention increase the activation of the intrinsic foot muscles potentially improving neuromuscular control. Gooding et al demonstrated using MRI that the exercises increase activation in all of the plantar intrinsic muscles by 9%-35%, depending on the muscle and the exercise (Gooding, Feger et al. 2016). There is limited definitive research determining whether improved strength is correlated with improved proprioceptive acuity; however some authors have recently reported improvements in shoulder joint position sense following a strength training intervention (Salles, Velasques et al. 2015). Their data showed significant improvements in mid-range joint-position reproduction following 8 weeks of strength training done at the same intensity. This was attributed to increased sensitivity of the muscles spindles following repetitive activation during the strength training program. Muscle spindles are under constant sensitivity regulation via gamma motoneuronal activity in order to remain sensitive to extrafusal fibre length changes during muscle contraction. Athletes and individuals regularly participating in exercise training have been shown to demonstrate better proprioceptive acuity than non-athletes/non-exercisers (Muaidi, Nicholson et al. 2009, Courtney, Rine et al. 2013) suggesting that the proprioceptive system itself can be improved with training. It must be stated that the findings from the aforementioned study (Salles, Velasques et al. 2015) are focussed on the shoulders and hence may not apply to the feet. Joint position sense is particularly important in the upper limbs in order to control reaching movements and complete tasks whereas the lower limb is primarily concerned with support and ambulation hence a similar degree of improvement may not directly translate. However the training protocol would serve as a new stimulus to the muscles within the feet increasing neural drive both efferent and afferent and thus it could be possible that alongside the muscle strength improvements witnessed in this study, there were potentially improvements in proprioceptive sensibility within these muscles following the 6 weeks of training. As the plantar intrinsic muscles have been shown to be involved in upright balance control (Kelly, Kuitunen et al. 2012), this improvement in proprioceptive acuity in these muscles could account for the improvements in static balance measures observed in this study, highlighting the role of plantar intrinsic foot muscle training in enhancing balance of older adults.

In contrast to the improvements witnessed in the balance parameters, the increase in foot strength appeared to have no effect on the gait performance variables assessed in this study. This has also been reported in a prior study on a similar age group investigating the effect of a foot gymnastics program in addition to a conventional whole body training program with no supplementary effect of foot training being observed (Hartmann, Murer et al. 2009). It could therefore be that the foot strength increases witnessed here may not have been detectable through the overall gait performance variables measured here with their effects primarily detectable through other dynamic tasks (Goldmann, Sanno et al. 2012) or direct force or muscle analyses used previously (Kelly, Lichtwark et al. 2015, Peter, Hegyi et al. 2015).

Following discussion of the results, the limitations associated with this research study and its outcomes should be outlined. The assessment of foot strength was limited to hallux plantar flexor strength as an indicator for foot strength therefore it is not known if improvements in lesser toe strength or toe abductor strength occurred and consequently their relative contribution to the associated balance improvements. We chose to assess purely hallux strength due to the greatest relative contribution for balance and gait tasks directed through the hallux in comparison to the lesser toes (Jacob 2001, Chou, Cheng et al. 2009, Tanaka, Hashimoto et al. 1996) and also the ability to maintain consistency of measurement through the isolation of one digit. Previous study has outlined the difficulty in recording the lesser toes with a dynamometer due to anatomical differences in foot shape (Goldmann and Bruggemann 2012). It should also be stated that the participants in this study were healthy and physically active and thus exhibited a good level of physical mobility and performance. Addressing the benefits of foot exercise training in a less mobile population and consequently at greater risk of falls could therefore be of interest.

Conclusion

In conclusion, the results from this study indicate that 6 weeks of home-based foot strength training can be effective at promoting increased foot strength in healthy older adults. Our findings provide additional support for the relationship between increased foot strength and a greater functional base of support. Foot strength training also leads to reductions in postural sway, possibly due to a concurrent improvement in proprioceptive acuity and neuromuscular control through the repetitive muscle activation with the exercises. With the high rate of adherence to this training program, similar interventions can be used in patients with foot muscle weakness to reduce their risk of falls.

Chapter 7

Does minimalist footwear improve balance and foot strength in older adults?

Simon Franklin, Michael J. Grey, François-Xavier Li

Abstract

Loss of balance is a common problem in older age and footwear is one factor which can have a considerable influence on stability and the increased risk of falls. Foot pain and problems are also common with approximately 25% of over 45's suffering from foot pain. Inadequate footwear fit and constricting structural designs is suggested to play a part in this foot pain prevalence. Minimalist footwear is reported to allow for natural foot motion and shape, reduce sensory interference and potentially increase activation of intrinsic foot muscles. The aim of this study was to explore if lifestyle use of minimalist footwear would improve the foot strength and balance of older adults. Thirteen healthy older adults (71years±4.0, 25.1±4.4 BMI) wore minimalist footwear in their daily life for a 4 month period whilst twelve controls (72years±3.1, 23.1±3.4 BMI) continued to wear their conventional footwear for the same period of time. Foot strength, static balance, dynamic balance and gait were assessed before and after the intervention and daily step count and minimalist footwear use were recorded for 1 week intervals within each month. In the intervention group there was a significant increase in hallux plantar flexor strength of the left foot and functional reach test performance whereas no significant changes were witnessed in the control group. No significant changes in either group were witnessed in static balance or gait measures. Twelve out of 13 participants in the intervention group reported an improvement in balance confidence whilst there were also reports of reductions in foot and hip joint pain with the use of minimalist footwear. The findings of this study show promise for the use of minimalist footwear in an older age population for the

improvement of foot muscle strength, balance confidence and the reduction of foot and joint pain.

Introduction

Ageing causes declines in the physical, sensory and cognitive mechanisms which contribute to an increased susceptibility to falls in older age. The propensity is highlighted by the statistic that one in three over 65's have at least one fall per year with this rising to one in two in the over 80's (NICE 2013). With this in mind there are multiple fall prevention strategies targeted at reducing this age related decline such as muscle strengthening and balance training programs as well as providing assistive devices (NICE 2013). One factor which is often considered as an avenue to improve balance is footwear. One in four older adults suffer from foot pain which impairs mobility and balance and inappropriate footwear can contribute towards this problem (Menz 2016). It has been widely reported, particularly in older adult females, that the shoes worn are too narrow in the toe box to accommodate the natural foot shape (Menz and Morris 2005). This constricting toe box volume leads to cramping of the toes and joint pathologies, increased dorsal and plantar pressures and foot pain symptoms (Paiva de Castro, Rebelatto et al. 2010, Branthwaite, Chockalingam et al. 2013). There is also suggestion that these constrictions imparted by the footwear structure can lead to a resultant weakening of important supportive foot muscles (Menz 2015).

Intrinsic foot strength was observed in the previous chapter (Chapter 6), and in prior research (Mickle, Caputi et al. 2016), to be positively associated with balance performance in older adults. Following a 12 week progressive resistance exercise program toe flexor strength significantly increased accompanied by increased single leg balance time (Mickle, Caputi et al. 2016). It therefore seems that maintaining strong foot function has important balance implications and ensuring that the footwear worn does not confine and constrict these functions is of paramount importance.

Minimalist footwear has been defined as "footwear providing minimal interference with the natural movement of the foot due to its high flexibility, low heel to toe drop, weight and stack height, and the absence of motion control and stability devices"(Esculier, Dubois et al. 2015). It is marketed as a method to mimic the way we walk barefooted allowing the foot to spread under load, distributing pressure more evenly across the foot (D'AoÛt, Pataky et al. 2009). Particularly for people suffering from foot problems such as hallux valgus this footwear may also serve to reduce the pressure placed on these particular points which may occur in conventional footwear with narrow toe boxes (Branthwaite, Chockalingam et al. 2013). Furthermore, via the thin flat sole, input interference to the sensory mechanisms may be reduced and the foot and lower leg musculature may be activated to a greater extent as the balance modulation reliance is placed back on the foot rather than on the footwear's built-in supportive structures (McKeon, Hertel et al. 2015, Franklin, M.J. et al. 2018). Previous research supports this suggestion with the finding that after running in minimalist shoes for 6 months there was a significant increase in foot and leg muscle volume whereas there was no change in the control group who continued running in standard running shoes (Chen, Sze et al. 2016). Similar muscle volume findings in addition to an increase in longitudinal arch stiffness have also been shown after 12 weeks of training in minimalist shoes (Miller, Whitcome et al. 2014). Research is now required to determine if simply walking in minimalist shoes can have a similar benefit to foot strength and if the balance of older adults can also be improved as a consequence.

Much research has examined the differences between walking in common footwear and walking barefooted (Franklin, Grey et al. 2015) and the previous chapters of this thesis (Chapter 3 and Chapter 4) explain how minimalist footwear fits into the equation. Much of this research however examines the acute exposure to walking barefooted or in minimalist footwear in the habitually conventional shod or how habitual shod populations directly compare to habitual barefoot populations. However, little research has focussed on the longitudinal effects of walking barefooted or in minimal footwear in habitual conventionally shod populations particularly in older age whom have many years of conventional footwear use.

Therefore the aim of this study was to investigate if wearing minimalist shoes in day-to-day life for a 4 month period would lead to improvements in balance in older adults through increased intrinsic foot strength and increased input to the sensory mechanisms. We hypothesised that we would observe an increase in hallux flexor strength and this would correspond to an improvement in functional reach test performance and a reduction in postural sway. We also hypothesised that we would observe an improvement in gait stability following the 4 month period as determined by a reduction in gait variability.

Methods

28 healthy and habitually physically active older adults (15 in intervention group: 4male, 71±3.9years, 24.68±4.23 BMI; 13 in control group: 3male, 71.8±3.7years, 23.28±3.35 BMI) volunteered. All participants visited the laboratory on two occasions; at baseline and for a 4-month follow up visit. Prior to commencing all participants completed a general health screening questionnaire and signed an informed consent as per the University ethical committee procedure. During both visits participants completed identical tests of static and dynamic balance, gait performance and foot strength. All measurements were undertaken by the same experimenter and at the beginning of each testing session their height and weight was measured whilst barefoot and lightly clothed.

The following tests were used to assess for dynamic and static balance. . The first test consisted of the functional reach test (FRT), a commonly used test of dynamic balance and margins of stability (Duncan, Weiner et al. 1990, Martins, de Menezes et al. 2012). Participants completed 3 trials on each side of the body and an average calculated. Following this, a postural sway task was completed on a force plate with feet in parallel stance, medial edges of their feet touching, hands on their hips and looking directly forwards. When comfortable, participants were instructed to close their eyes and keep them closed for the duration of the trial and to try and remain standing as still as possible. The eyes were closed in order to remove the overriding influence of vision, increasing the difficulty of the task by causing the participants to solely utilise their remaining sensory modalities including proprioception and sensory feedback from the foot soles. Each trial began as soon as the eyes closed, lasted for 35seconds and was repeated 3 times. The first 5 seconds of each trial was discarded in order to remove any initial destabilisation effects as a result of the sudden loss of vision.

The force plate (Kistler Type 9281C, dimensions 600x400mm) recorded at a sampling rate of 1000Hz and using inbuilt Bioware software algorithms the centre of pressure (COP) distance in the anterior-posterior (AP) and medial-lateral (ML) directions was calculated from the force. The resultant distance (Res) was then calculated using the AP and ML distances. Each trial was low pass filtered using a zero-lag fourth order Butterworth filter with a cut off frequency of 8Hz. Average displacement, range of the displacement and sway velocity were calculated for the ML, AP and Res directions for each trial and an average of the 3 trials taken. Sway path length was calculated by summing the absolute Res distance from the initial starting position. Sway area was approximated using principal component analysis (PCA) by surrounding the sway path with an ellipse and a bounding box and calculating the area of the ellipse as an approximation of the sway area. The PCA involved calculating the eigenvalues and eigenvectors from the covariant matrix using all the ML and AP data points and their mean values. The data points were orientated such that they were centred about the origin and were rotated by the angle of the first eigenvector. The maximum and minimum values of the transformed data set were then found to determine the size of the bounding box with the eigenvectors specifying the axes. The prediction ellipse was then fitted using the same eigenvectors specifying the main axes of the ellipse and the area of the fitted ellipse calculated as an estimation of the sway area.

To assess gait participants completed a 10m walk test barefooted and in a pair of standardised shoes (Style Code: 10001, Hobos Womens, Style Code: 50109, Hobos Mens). Retro-reflective kinematic markers were placed according to the Vicon plug-in gait lower body template. A 13 camera Vicon MX system (Vicon, Oxford, UK) recording at a sampling rate of 100Hz was used with calibration prior to data collection completed such that there was a residual error of less than 1mm. Participants completed 6 trials in each footwear condition; 3 at their self selected preferred walking speed and 3 at their fastest walking speed. There was a pre-determined mark placed on the floor from which each participant started which allowed for a minimum of 3 steps prior to reaching the recording field of view. Participants walked from this point to another pre-determined mark on the floor, placed outside of the recording field of view before returning to starting mark and repeating. The recording field of view captured the middle 6m which was used for analysis.

Gait speed was calculated using the time taken for the right hip marker (RASI) to cover the middle 6m. The heel and toe markers (R/LHEE, R/LTOE) were used to determine heel contact and toe off events. Gait variability, namely stance time variability and step width variability from step to step, calculated here as the percentage coefficient of variation ((standard deviation/mean)*100), was used as an indicator of gait stability due to its high validity previously reported in a recent review (Bruijn, Meijer et al. 2013). All data analysis was completed using custom-written scripts in Matlab (MATLAB, The MathWorks, Natick, MA, USA).

The final test assessed the plantar flexor strength of the hallux. This was completed using a custom built device previously tested for reliability. The foot was placed onto the base plate with the ankle positioned into 35° plantar flexion and the hallux in a position of 35° dorsiflexion. This position was chosen to simulate the role of the hallux during the final point of toe off during walking whilst limiting the external influence of larger ankle plantar flexors. The foot was strapped down and the participant was instructed to maintain this foot and leg position, sitting upright with their arms across their chest. The task involved isometric hallux plantar flexion onto the wedge affixed to the strain-gauge (NeuroLog NL62). The raw data was amplified and digitized using an analog-to-digital card (National Instruments BNC-2090A, National Instruments USB-6251) and the data sampled by Mr Kick III software for Windows 7. Three sub-maximal practice trials were given to each participant and feedback provided by the experimenter if required. A set of three recorded maximum voluntary isometric contractions (MVIC) lasting 3-5seconds followed with a recovery period of 1 minute in between repetitions. The same procedure was then completed with the other foot. The peak force was computed and normalised with respect to bodyweight (%BW) before being used in the analysis.

These testing protocols were completed at baseline and then repeated at the 4 month follow up session.

Statistical analysis was completed using SPSS V. 22.0 for Windows (IBM Corporation, Somers, NY). All variables from baseline to follow up were compared using paired samples t-tests with levels of significance set to $p<0.05$.

Further data was collected during the intervention period. Participants were provided with a diary and a pedometer (Yamax CW-600) and were instructed to document their footwear use (time spent wearing the minimalist shoes) as well as their habitual activity (step count) for random 1 week intervals within each month. In terms of the footwear use participants were instructed to write the time in the diary each occasion they put the minimalist shoes on and subsequently when they took them off again for each separate period throughout the day to enable us to understand how much time was spent wearing the shoes. Participants were also instructed to wear a pedometer for the duration of the day regardless of whether they were wearing the minimalist shoes. At the end of each day they were instructed to fill in the diary with the total number of steps completed to provide us with data on their overall habitual activity.

At the end of the 4 month intervention period and prior to the follow-up testing session, qualitative data was also collected from all participants to understand their opinions of the shoes and if they experienced any changes in balance or confidence over the 4 month period. This was completed in the form of a 7-item questionnaire with a mix of open ended, likert scaled and dichotomous questions. The questions assessed what the participant's "liked and disliked the most about the minimalist footwear" (open ended); if the participants "felt their balance had changed" after the 4months of wearing the minimalist footwear (5point scale; 1=Definitely Worse, 5 = Definite Improvement); if they "felt more confident in their balance wearing minimalist footwear as opposed to conventional footwear"(5point scale; 1=Not at all, 5 = Definitely); if they were going to continue wearing minimalist footwear (5 point scale; 1=Never, 5=Always) and if they would recommend this type of shoes to other people in their age group (yes or no).

Results

Group	n	Age (years)	BMI (kg/m^2)		FRT (cm)			
			Baseline	Follow Up	Baseline		Follow Up	
					Right	Left	Right	Left
Intervention	13	71±4.0	25.1±4.4	25.1±4.3	31±7	30±7	34±4*	34±4*
Control	12	72±3.1	23.1±3.4	23.0±3.2	31±4	30±5	29±5	30±5

Table 12 - A summary of the group characteristics and their respective functional reach test performances (FRT) at baseline and follow up. FRT is displayed for both the right and left side of the body. Data displayed is means ± standard deviation. BMI = body mass index. FRT = functional reach test performance. * indicates a significant difference from baseline to follow up (p<0.05).

Two participants dropped out of the intervention group and one participant from the control group. Reasons for drop out from participants in the intervention group included not being able to become accustomed to flat, unsupportive nature of the minimalist footwear and the inability to get correct size to comfortably fit foot shape for each participant respectively. The reason for dropout for the participant from the control group was a back complaint and subsequent inability to complete normal habitual activity. The following data is for the remaining participants as displayed in Table 12.

Participants in both groups showed a good level of physical activity with an average of 8360±4830 steps per day for the intervention group and 8033±2850 steps per day for the control group across the 4 months. The intervention group on average spent 414±151 minutes per day wearing the minimalist shoes whilst the control group wore conventional footwear for an average of 715±233 minutes per day.

There was a significant increase in functional reach test performance from baseline to follow up (Right: $t(12)=2.153, p=0.052$; Left: $t(12)=2.597$, $p=0.023$) in the intervention group but not in the control group (Right: $t(12)=0.967, p=0.354$; Left: $t(11)=2.014$, $p=0.069$) as shown in Table 12.

In the intervention group there was an increase in hallux plantar flexor peak force from baseline (M=5.5%BW, SD=2.7) to follow up (M=7.3%, SD=3.2); $t(12)=4.232$, $p=0.001$ in the left foot but not in the right foot (baseline: M=7.5%BW, SD=3.2; follow up: M=7.8%, SD=4.8; $t(12)=0.293$, $p=0.774$). There were no significant changes in the control group in the left foot (baseline: M=5.7%BW, SD=2.3; follow up: M=6.3%BW, SD=3.0; $t(11)=1.062$, $p=0.311$) or right foot (baseline: M=7.2%BW, SD=1.9; follow up: M=7.6%BW, SD=3.0; $t(11)=0.673$, $p=0.515$) (Figure 23).

Whilst there were differences observed between walking barefoot and walking in conventional footwear in both groups at baseline *(for example preferred gait speed: Intervention group: Barefoot: M=1.44, SD=0.17 vs Shoes: M=1.56, SD=0.18; $t(12)=3.013$, $p=0.011$; Control Group: Barefoot: M:1.36, SD=0.17 vs Shoes: M=1.44, SD=0.12; $t(11)=2.432$, $p=0.033$)* and at follow up *(for example preferred gait speed: Intervention group: Barefoot: M=1.44, SD=0.14 vs Shoes: M=1.57, SD=0.17; $t(12)=5.505$, $p<0.001$; Control Group: Barefoot: M:1.42, SD=0.14 vs Shoes: M=1.52, SD=0.11; $t(11)=2.482$, $p=0.030$)* there were no differences in gait variables between walking barefoot/in conventional footwear at baseline to the same footwear conditions at follow up *(Intervention Group: Barefoot: Baseline: M=1.44, SD=0.17 vs Follow Up: M=1.44, SD=0.14; t(12)=0.019, p=0.985; Shoes: Baseline: M=1.56, SD=0.18 vs Follow Up: M=1.57, SD=0.17; t(12)=0.171, p=0.867, Control Group: Barefoot: Baseline: M=1.36, SD=0.17 vs Follow Up: M:1.42, SD=0.14; t(11)=2.005, p=0.070, Shoes: Baseline: M=1.44, SD=0.12 vs Follow Up: 1.52, SD=0.11; t(11)=2.029, p=0.067).*

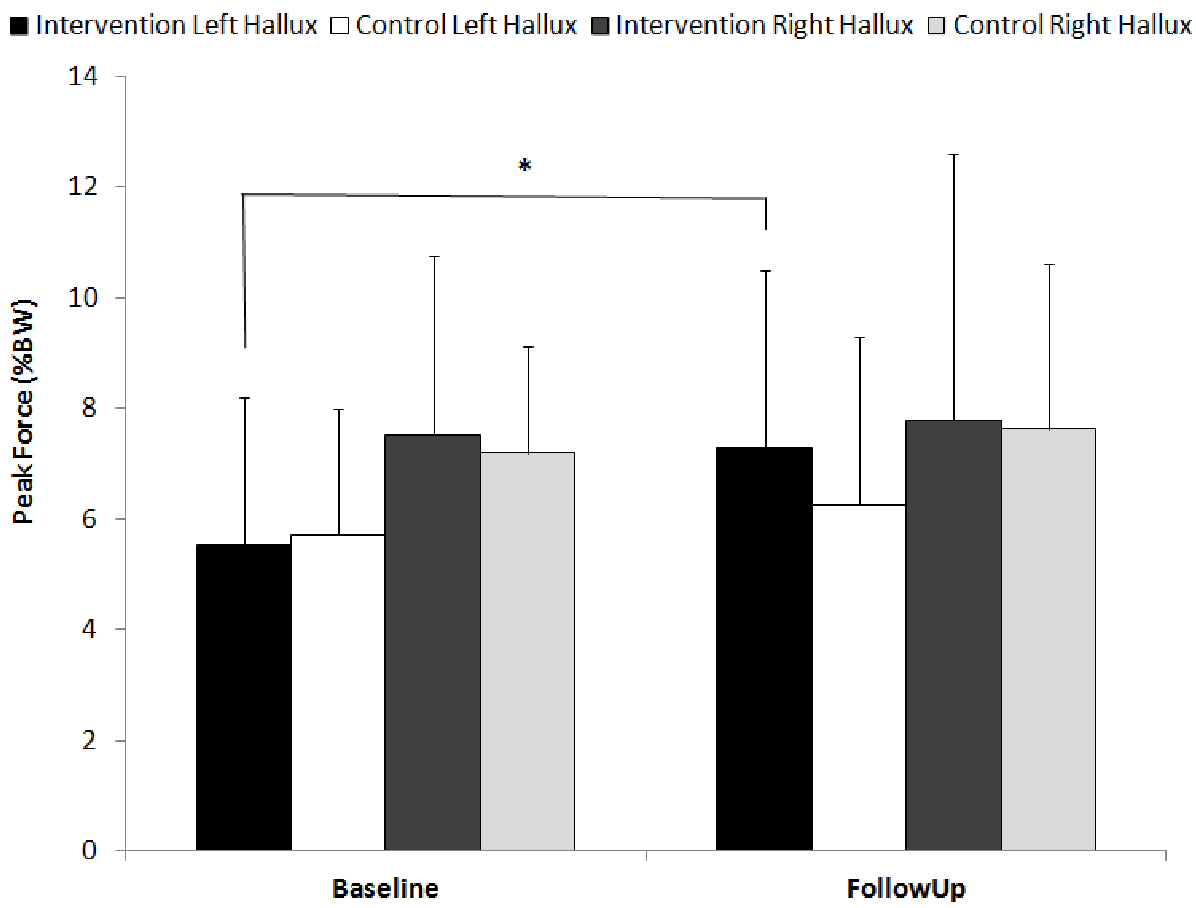


Figure 23 - Illustrating the change in peak hallux plantar flexor force as a percentage of body weight (%BW) from baseline to follow up. The black bars represent the average force in the left hallux in the intervention group. The white bars the average force in the left hallux in the control group. The dark grey bars the average force in the right hallux in the intervention group and the light grey bars the average force in the right hallux in the control group. Error bars indicate standard deviation across the group. * indicates a significant difference from baseline to follow up (p<0.05).

The same was true for step length with differences between footwear conditions at baseline and follow up *(Baseline: Intervention group: Barefoot: M=698, SD=66 vs Shoes: M=764, SD=72; t(12)=6.358, p<0.001; Control Group: Barefoot: M:664, SD=75 vs Shoes: M=712, SD=76; t(11)=3.578, P=0.004; Follow Up: Intervention group: Barefoot: M=704, SD=57 vs Shoes: M=778,* *SD=68; t(12)=10.057, p<0.001; Control Group: Barefoot: M:677, SD=75 vs Shoes: M=737, SD=73; t(11)=5.038, P<0.001)* but no differences in the same footwear conditions between baseline and follow up *(Intervention Group: Barefoot: Baseline: M=699, SD=66 vs Follow Up: M=704, SD=57; t(12)=0.417, p=0.684; Shoes: Baseline: M=764, SD=72 vs Follow Up: M=778, SD=68; t(12)=1.467, p=0.168, Control Group: Barefoot: Baseline: M=664, SD=75 vs Follow Up: M=677, SD=75; t(11)=1.424, p=0.182, Shoes: Baseline: M=712, SD=76 vs Follow Up: M=737, SD=73; t(11)=2.482, p=0.030)*. Step width, stance time variability and step width variability showed no differences between footwear conditions within the testing time points or from baseline to follow-up following the 4 month period of minimalist shoe use. The control group did display a significant increase in fastest walking speed and concomitant reduction in stance time from baseline to follow up when walking barefoot but no significant changes in other gait measures or gait stability variables.

12 out of the 13 participants in the intervention group reported an improvement in confidence in their balance when wearing the minimalist shoes with 8 of those indicating that they felt their balance itself had also improved as a result. One participant stated:

"(I'm) more confident in walking in them, I sometimes used to trip in other shoes but so far never in the minimalist footwear".

Primary themes that participants gave for the improved confidence and balance was greater awareness of surface changes, greater feel of the ground as well as improved weight distribution across the foot due to the flat and wide shoe structure. Eight of the 13 participants stated they will regularly continue wearing these shoes with the remaining 5 stating they will wear minimalist shoes sometimes, primarily dependent on the weather. Of great interest is that participants also noted improvements in long-standing foot and joint pain following the introduction of wearing minimalist footwear. A second participant described:

"It's been a revelation to me, reduced foot pain from day of walking/touristing to a low level or zero...there is less seam pressure on arthritic joint... foot could spread fully"

Another quoted:

"I don't have pain in my hips since using the footwear – I have increased my walking".

A further participant also stated an improvement in toe feeling and movement during the minimalist footwear intervention:

"2 toes that 'didn't work' started to be able to move after about a month of wearing the shoes".

Discussion

This study set out to determine if wearing minimalist shoes in daily life might improve the balance of older adults. It aimed to assess if the lack of supportive features and high flexibility principal to minimalist footwear would enhance intrinsic foot strength through increased activation and reliance on the foot musculature. Furthermore we aimed to explore if as a result of the thin sole and the reduction in input interference, sensory sensitivity may be improved presenting as improvements in postural balance and gait stability. The results for these areas of discussion appear inconclusive as whilst the left foot of the intervention group displayed a significant improvement in strength the right foot was comparable to the control group with no significant increase witnessed. Likewise there was a significant improvement in functional reach test performance in the intervention group, coinciding with the improvement in strength in the left foot, however there were no significant improvements in either group in postural balance or gait performance. On the other hand, the qualitative data collected from the participants in the intervention does clearly suggest the minimalist footwear has had a beneficial influence over and above purely a balance perspective with motivating reports of a reduction in long-standing joint/foot problems and pain and a resultant increase in activity.

Firstly the participant's responses to the questionnaire highlighted potential benefits to balance and confidence with the main theme being repeatedly stated as the reason for improvement being greater awareness of the surface/greater feel of the ground.
This theme points towards our hypothesis that due to the thin sole of the minimalist footwear the inputs to the sensory mechanisms may be heightened. Whilst we observed no improvement in postural balance or gait stability measures in the quantitative data, seven of the participants stated the increased feel of the ground or awareness of surface changes experienced with the minimalist footwear as the primary reason for their improved balance confidence. This qualitative data suggests that sensation from the foot sole is of prime importance to stability. It would therefore be recommended that in future study on this topic sensory sensitivity could be included in the assessment measures to ascertain if improvements in this sensory modality are witnessed and if these correspond to improvements in balance and stability. Common ways to measure sensory sensitivity of the foot sole are through vibration or monofilament testing. Vibration can be used to perturb the sensory receptors as they are stimulated by the vibration and elicit a firing rate response. By varying the size of the stimulus and the frequency, the perceptual threshold can be determined and used as an indicator of sensory sensitivity. Monofilaments can also be used to determine the perceptual threshold whereby you apply a range of monofilament sizes which vary in force to the foot sole and the participant is required to respond with whether they can perceive the filament or not. The threshold is then established as the lowest monofilament which can be correctly perceived a pre-determined percentage of the trials. These assessments should be done at different sites across the foot sole to assess areas with different skin hardness, thickness and receptor location.

Previous research has demonstrated this importance of the plantar cutaneous afferents for maintaining gait stability (Lin and Yang 2011). In this study participants were required to take a step and begin walking from their standing position on a force platform with and without plantar cutaneous afferent desensitisation from cold water immersion of the stepping foot, supporting foot or both feet. Following desensitisation the medial lateral (ML) centre of pressure (COP) displacement towards the stepping leg was significantly reduced during the anticipatory phase of step initiation with the greatest reduction being when both feet were desensitised (Lin and Yang 2011). This points towards the role of the cutaneous afferents in modifying anticipatory postural adjustments (APA's) and compensatory postural adjustments (CPA's) based on limb loading. APA's are concerned with the activation or inhibition of postural and support muscles (trunk and leg) to negate the potential disruption to balance through a predicted perturbation (Santos, Kanekar et al. 2010). They are vital in preparing the body for movement or for a predicted perturbation. CPA's also activate/inhibit the postural and support muscles but serve to restore stability once the perturbation/movement has arisen based on information provided by sensory feedback signals(Park, Horak et al. 2004). This highlights how maximising plantar cutaneous sensitivity could be vital to ensure appropriate postural stability during step execution (Lin and Yang 2011). This is particularly relevant in the older ages who, despite having reduced cutaneous sensitivity (Wickremaratchi and Llewelyn 2006), have been shown to rely more on foot sensitivity for postural control than their younger counterparts (Machado, da Silva et al. 2017). More recently a study replicated the stepping task but instead of plantar desensitisation they simply had participants barefooted or wearing shoes (Vieira, Sacco Ide et al. 2015). The results showed that wearing shoes as opposed to being barefoot evoked a similar reduction in ML COP displacement symptomatic of cutaneous afferent interference and an altering of APA's (Vieira, Sacco Ide et al. 2015). The participants in this study indicated an increased awareness/greater feel of the ground when wearing the minimalist footwear appearing to advocate that this footwear may be effective at reducing the cutaneous afferent interference and subsequent APA adaptation associated with conventional footwear, however further research is required to determine this.

An explanation as to why we saw no improvement in the gait or the gait stability variables could be that we completed these baseline and follow up tests barefoot or in a pair of standardised conventional shoes and not in the minimalist footwear given to the participants for the intervention. Whilst participants reported improvements wearing minimalist footwear this was in comparison to their previous conventional footwear worn. It therefore seems that wearing minimalist footwear for 4 months does not lead to any lasting changes in how we walk when returning to conventional footwear (i.e. it doesn't lead to adopting a more similar gait pattern to barefoot) and that we do revert back to the gait kinematics attributable to that footwear condition. It is possible that a considerably longer amount of time is required to see perennial changes in gait kinematics or that simply wearing conventional footwear forces a different gait pattern to be adopted. Future studies should complete the baseline and follow up assessments in the minimalist footwear worn throughout the intervention to observe the transition period to this type of footwear and elucidate any changes in gait kinematics.

The significant changes which were witnessed in the control group from baseline to follow up were isolated to purely the fastest walking speed (3.28% increase in fastest walking speed and 4.19% decrease in stance time during the fastest walking speed condition) and are likely attributable to participant/experimental error in self-selecting their walking speed rather than a meaningful finding. That there were no differences in preferred walking speed from baseline to follow up in this group further supports this suggestion of an anomalous finding.

Similarly, at both testing sessions postural sway was measured barefoot, with no changes witnessed; this therefore suggests that no perennial enhancement in the sensitivity of the sensory mechanisms occurred over the 4 month period in order to evoke perceptible changes in the follow up assessment. Alternatively, the qualitative data indicates that sensory sensitivity and subsequently balance and confidence may be improved, in an acute state, by replacing conventional footwear with minimalist footwear due to the reduction in foot-floor interference due to the thin, flat and flexible sole. It is not known if longer term use of minimalist footwear would lead to lasting improvements.

An interesting finding from the qualitative data, which was not originally expected but should be explored, is the reports of clear improvements in long-standing joint pains or foot problems following the introduction of the minimalist footwear. The first and probably least surprising is the aforementioned report of “reduced foot pain... less pressure on arthritic joint...foot could spread fully”. The type of minimalist shoe used in this study was designed to be foot shaped and thus offered a wider toe box than many conventional footwear options. The reports of “reduced foot pain” and “foot could spread fully” highlights the benefit of a wide shoe structure in the forefoot which allows the foot to spread fully under load and relieve pressure. This is supported by previous research describing reduced forefoot widening when walking in conventional shoes as opposed to walking barefoot suggesting a form of constriction to the natural movement of the foot (Wolf, Simon et al. 2008). A common issue in older age is hallux valgus with a recent review estimating over 1 in 3 over 65’s suffer from the problem (Nix, Smith et al. 2010). This along with other arthritic conditions can lead to increased width at the forefoot and/or unnatural shape causing issues of localised pressure points and pain often prominent when wearing footwear (Paiva de Castro, Rebelatto et al. 2010, Branthwaite, Chockalingam et al. 2013). With this in mind, in a study on 176 older adults, it was found that 81.4% wore indoor shoes and 78.4% wore outdoor shoes narrower than their foot width clearly highlighting the issue addressed (Menz and Morris 2005). Hence wearing a shoe wide enough to satisfy not placing undue pressure on potentially painful areas can be especially noteworthy for this population and providing shoe options, like the one used in this study, could be particularly relevant for this population.
Also reported was the presence of hip pain, previously experienced regularly during walking, had subsided when wearing minimalist footwear. This allowed this participant to increase the

amount they could walk supported by an observed 15% increase in step count from the first month to the fourth month. There has been previous evidence that walking barefoot or in flat, flexible shoes can reduce hip and knee joint loading (Shakoor and Block 2006, Kemp, Crossley et al. 2008, Shakoor, Sengupta et al. 2010) and also how this type of shoe can reduce the pain experienced and improve function in osteoarthritis sufferers (Trombini-Souza, Kimura et al. 2011). The main change witnessed in these footwear conditions was a reduction in the knee adduction moment, used as a measure of force applied through the medial compartment of the knee, which has previously been associated with the progression of knee osteoarthritis (Foroughi, Smith et al. 2009). A reduction in hip adduction, internal and external rotation were also witnessed when walking barefoot as opposed to wearing walking shoes (Shakoor and Block 2006). These reductions in loading can reduce the risk of joint inflammatory responses, bone hypertrophy and related pain symptomatic of osteoarthritis (Aresti, Kassam et al. 2016). It is promising to hear first hand from a participant that minimalist shoes have successfully appeared to reduce the loading at the hip and subsequent pain within a 4 month period.

A third participant described regaining movement in 2 toes which they had previously been unable to move. This is purely anecdotal evidence however it could be the consequence of increased activation of intrinsic foot muscles previously implied when barefooted or in minimalist footwear (Miller, Whitcome et al. 2014, McKeon, Hertel et al. 2015, Chen, Sze et al. 2016) supported by a clear increase in foot strength witnessed in this participant from baseline (right foot: 4.1%BW, left foot: 4.6%BW) to follow up (right foot: 6.0%BW, left foot: 7.6%BW). Muscle size and strength is known to reduce with ageing even in the intrinsic foot muscles (Mickle, Angin et al. 2016) and this can have implications on foot and whole body functions such as arch support (Kelly, Cresswell et al. 2014), efficient force transmission during walking (Kelly, Lichtwark et al. 2015) and balance (Spink, Fotoohabadi et al. 2011). Foot strength improvements were observed in this study alongside corresponding increases in functional reach distance.

These findings coupled with the reports of improved movement of the toes support the suggestion that minimalist footwear may be beneficial at increasing the activation of intrinsic foot muscles and helping to maintain the function of these muscles.

Whilst the findings of this study show promise for future research into minimalist footwear use in older adults the following limitations should be taken into consideration. Whilst care was taken to remain neutral and ensure the study was portrayed as purely exploratory, due to the nature of the study and the requirement of changing footwear habits for an extended period of time it is possible that the intervention sample had a heightened interest in the footwear and study from a personal point of view. It is possible that these participants represent a population who struggle with finding suitable comfortable footwear and are looking for an alternative and therefore the findings may not be applicable to the older age population as a whole. Due to the diverse nature of foot shapes and the lasting effects of previous footwear habits, as highlighted by the dropouts in this study, it is clear that whilst minimalist footwear may be beneficial for some it may not be for others and as such should be determined on a case by case basis.

Conclusion

In conclusion, the tests of balance and gait used in this study did not detect any perennial changes in postural balance and/or gait stability following 4 months of wearing a minimalist shoe. Conversely, the qualitative data collected from participants appears to indicate some perceptible improvements in balance and balance confidence when walking in minimalist shoes in comparison to conventional footwear predominantly as a result of the thin, flexible sole and wide fit. Furthermore, aside from our original research question, direct feedback from participants appears to indicate that wearing minimalist footwear may be an effective way at reducing the pain experienced through osteoathritic problems in the feet and hip joints. These reports prompt the need for more research over a longer period and in different age populations to determine if lasting improvements in foot strength, balance and gait as well as reducing foot/joint pain can be observed with minimalist footwear use.

Chapter 8 – General Discussion

It is well established that within the ever growing older age population the occurrence of falls and fall related injuries is a substantial problem for the individual, the health service and associated support networks. As such there is an abundance of research focussed on optimising the balance of older adults and maintaining mobility through a range of avenues including physiological, psychological and nutritional interventions, additional support and assistive devices as well as social and environmental initiatives. The focus of this research was to explore one factor which could affect balance and play a role in the susceptibility to falls: footwear.

The foot provides the only contact point between the body and the ground. As such a change at this contact point, through the inclusion of footwear, can have a considerable influence on stability through not only interference between the foot and the ground but also a potential modification to the alignment of the foot, leg and body as a result of the footwear structure (Menz and Lord 1999). It has already been shown previously how inappropriate footwear, such as; having an insufficient fit (Menz and Morris 2005, Menant, Steele et al. 2008), comprising a heel (Lord and Bashford 1996, Tencer, Koepsell et al. 2004) or thick, soft soles (Robbins, Waked et al. 1994, Robbins, Waked et al. 1995, Robbins, Waked et al. 1997) can all contribute to impairment in balance and an increase in the risk of falls. The role of minimalist footwear, specifically designed to reduce the interference between the foot and the ground and maintain natural foot, lower leg and body alignment, was examined in this thesis to determine if there were benefits to the older adult population in terms of balance and gait performance. The purpose of this thesis was to:

i. Determine if walking in minimalist footwear reduces the interference and changes witnessed when walking in conventional footwear and if it is a viable alternative to walking barefoot.

ii. Improve the understanding of whether wearing minimalist footwear is beneficial for older adults to wear to improve their balance and reduce their risk of falls.

The first experimental chapter (Chapter 2) served to review the current literature to establish how common footwear may affect gait. It highlights how walking in conventional footwear causes changes in the kinematics and kinetics of gait. When people walk barefooted they tend to walk at a slower speed, take shorter steps, contact the ground with a more plantar flexed ankle, have greater knee flexion at foot ground contact and display greater degrees of foot motion. Likewise habitual barefoot populations exhibit considerably wider feet, more consistent arch heights and an improved plantar pressure distribution across the foot sole when walking barefooted compared to a habitually shod population. Kinetic differences were also witnessed with a reduction in vertical ground reaction force as well as reduced knee (varus and flexor) and hip (flexor and extensor) moments when walking barefoot however the data on these findings was less abundant and hence less conclusive. Similarly it highlighted the paucity of research on the effects of footwear on muscle activity, the lack of research investigating minimalist footwear and also the dearth of understanding on the differences between walking barefoot and in footwear in the older adult population.

Chapter 3 and Chapter 4 attempted to fill these knowledge gaps by investigating how minimalist footwear fits into the equation between barefoot and conventional footwear in terms of kinematic, kinetic and muscle activity differences during walking across a wide age range including the older ages. Chapter 3 focussed on the kinematics and kinetics and determined that minimalist footwear was intermediate to barefoot and conventional footwear. Walking barefoot resulted in reduced step and stride lengths and increased ankle plantar flexion at foot-ground contact across all age groups whilst only the middle and older ages also exhibited a reduced gait speed unshod compared to shod. Similarly all age groups exhibited shorter step and stride lengths, increased ankle plantar flexion and increased knee flexion at foot-ground contact when walking in minimalist footwear in comparison to conventional footwear. Although a reduction in peak loading ground reaction force was witnessed when walking barefooted, no corresponding reduction as a result of the kinematic changes was observed in the minimalist footwear. It appears that wearing minimalist footwear does not completely replicate walking barefooted but does reduce the kinematic changes associated with wearing conventional footwear offering the most viable alternative to walking barefoot. The reasons for minimalist footwear not completely replicating walking barefoot are still to be elucidated however it could be due to cutaneous afferent interference and/or the perception of protection with the addition of any structure encompassing the foot. Ultimately however it appears that regardless of age and years spent wearing conventional footwear the kinematic changes associated with walking barefoot or in minimalist footwear are consistent.

Chapter 4 aimed to determine if walking barefoot or in minimalist footwear leads to increased lower leg muscle activity in comparison to conventional footwear. Similarly to the kinematic findings described in Chapter 3 walking in minimalist footwear had more similar muscle activation patterns to walking barefoot than walking in conventional footwear. Walking barefoot or in minimalist shoes both resulted in a reduction in tibialis anterior activity at initial stance due to a flatter foot at contact and less eccentric loading. There was also a reduction in peroneus longus activity at initial stance in the barefoot and minimalist shoe conditions but only in the young and middle age groups. A reduction in peroneus longus activity suggests improved stability and thus could indicate improved proprioceptive sensibility through increased afferent information with reduced footwear interference. Age-related decrements in proprioceptive and cutaneous acuity could explain the lack of difference in peroneus longus activity in the older age groups due to an inability to take advantage of the increased afferent information available. The young and middle age groups also displayed a reduced gastrocnemius medialis activity during the single support phase when wearing conventional footwear which suggests a reduced reliance on this muscle for balance modulation. The lack of difference in the older ages could be due to a greater reduction in walking speed when barefoot offsetting the increased recruitment of the gastrocnemius medialis muscle. Nonetheless it appears that an increased activation of the gastrocnemius medialis when minimally/unshod could lead to strength improvements and improved proprioceptive sensibility over time. It must be emphasised however that this study was an acute exposure to minimalist footwear and thus these long term effects of walking in this footwear are unknown and thus remain purely speculative.

Chapter 5 reported a reliability study which aimed to assess if a custom built device was sufficiently accurate at assessing hallux plantar flexor strength across a range of joint angle combinations between two measurement time-points. The results from the study illustrated high repeatability between sets with correlation coefficient values ranging between 0.933 and 0.961 depending on the ankle and 1MPJ position combination and the non-dominant or dominant foot. This study therefore illustrated that this device and procedure was suitable to assess hallux plantar flexor strength in future studies across two time points.

Chapter 6 investigated the efficacy of a 6-week home-based seated foot exercise program at improving foot strength and balance in a sample of healthy older adults. It was concluded that this 6-week exercise program was effective at improving foot strength as evidenced by a 31% and 29% increase from baseline in hallux plantar flexor strength of the right and left feet respectively. This increase in foot strength was also associated with an improvement in functional reach test performance and a reduction in postural sway during a static balance task. This study further reinforces the link between hallux plantar flexor strength and the functional anterior base of support. It also suggests that repetitive activation of intrinsic foot muscles through daily foot exercise training may contribute to improvements in postural balance potentially via associated proprioceptive sensibility improvements in the foot muscles.

Chapter 7 was concerned with whether daily use of minimalist footwear over an extended period could be a method by which to increase foot muscle strength, improve balance and gait stability in older adults. The results demonstrated that 4 months of wearing minimalist footwear for an average of 7 hours a day resulted in a 33% increase in foot strength in the left foot whilst the right foot remained relatively consistent (4% increase). Similar to the results witnessed in Chapter 6, this increase in strength was associated with an improvement in functional reach test performance but no changes in gait stability or on this occasion postural balance measures. Qualitative feedback from participants suggested that improvements in balance and balance confidence were experienced when wearing minimalist footwear with the thin sole and wide fit, allowing for greater feel of the ground and room for foot spreading, given as the reasons for the reported improvements. Participants also reported reductions in foot and joint pain following use of minimalist footwear supporting previous research suggesting walking barefoot can be beneficial for reducing joint loading and pain symptoms in sufferers of lower limb osteoarthritis. Overall this research study shows promise and highlights the need for future work investigating the benefits of minimalist footwear on foot strength and balance across various age populations as well as for reducing the pain and progression of osteoarthritic conditions in the foot and joints of the lower limb.

The qualitative findings from this study further support previous research demonstrating the positive benefits of minimalist footwear in knee osteoarthritis patients through decreasing knee loading and improving pain symptoms (Sacco, Trombini-Souza et al. 2012, Shakoor, Lidtke et al. 2013). Whilst benefits of minimalist footwear could exist for this population care should be taken with recommending this footwear for people with other conditions. It is apparent that minimalist footwear by nature offers considerably less cushioning and support for the wearer. This could mean that if appropriate kinematic alterations are not made, i.e. reduced dorsiflexion, shorter stride lengths, slower gait velocity, then increased impact forces could be experienced during gait. Whilst for the general population this is not of major concern, for patients suffering from osteoporosis with lower bone mineral density this could put them at greater risk of stress fractures and thus ensuring they have footwear which provides a degree of cushioning could be beneficial for this population. As such recommendations to the older age population needs to be done on a case by case basis depending on the various conditions which they may suffer from but in the case of knee osteoarthritis and foot problems such as hallux valgus and pes planus it would appear that wearing minimalist footwear could be of measurable benefit.

It should be recognised that the majority of the participants used in the studies comprising this thesis are fit and physically active and thus may not be representative of the general older adult population. As such the outcomes from studies may be perceived as small. It must therefore be acknowledged that greater benefits to balance, confidence and mobility may be witnessed in those with lower baseline ability. Consequently the reports of improved balance confidence with use of minimalist are especially promising given the high baseline physical activity level. As confidence can be a significant barrier to physical activity in older age this footwear change could provide an efficient solution to this issue particularly for those less able and thus research should be directed towards this population.

It must be acknowledged that the number of participants from which these conclusions are based is small and thus whilst showing much promise definitive conclusions on the benefits of minimalist footwear should be drawn following larger and more long term studies. Additionally the training and intervention studies solely assessed hallux plantar flexor strength as an indicator of foot strength and thus it would be interesting in future studies to assess other muscle across the foot to have a clearer indicator of the muscular changes within the foot.

This thesis has provided a strong background for the benefits of wearing minimalist footwear in everyday life. It has demonstrated that people, regardless of age, inherently walk differently barefoot compared to when wearing footwear such that they modify their gait (via increased ankle plantar flexion, increased knee flexion and shorter step lengths) to control their contact with the ground, suggesting a greater amount of care is taken. Walking in minimalist footwear also resulted in these modifications to gait compared to conventional footwear however not to the same extent as walking barefoot. This indicates that, although they can be seen as the most viable alternative, they are not entirely comparable. Although there are specific features required to be deemed a minimalist shoes, as outlined in the consensus definition by Esculier et al (Esculier, Dubois et al. 2015), it is clear that there is a substantial degree of difference across shoes marketed as minimalist footwear.

Figure 24 – Two examples of a minimalist shoe with identical heel to toe drop of 4mm but clear differences in stack height. A) Merrell Vapour Glove, B) Nike Free 3.0

These differences can be quantified through various features of footwear including the stack height, last shape and heel to toe drop height. For example if we compare two shoes which are both marketed as minimalist footwear; the Merrell Barefoot Vapour Glove (Figure 24A) and the Nike Free 3.0 (Figure 24B), they have considerably different stack height besides specifying the same heel to toe drop of 4mm. The Nike Free 3.0 offers a significant degree of cushioning through two layers of foam across the whole sole of the foot and thus although satisfying the high degree of flexibility criteria through its segmented midsole construction and its low heel to toe drop, this greater stack height means it is considerably different to the Merrell Barefoot Vapour Glove which only has a maximum stack height of 6mm and minimum of 2mm.

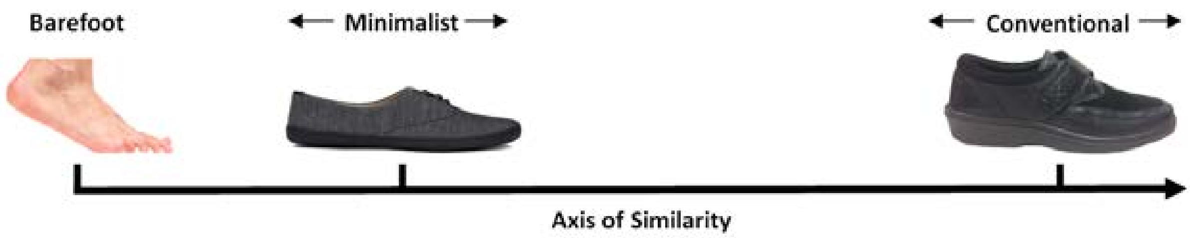


Figure 25 - Illustrating how minimalist footwear fit intermediately on a continuum of similarity between barefoot and conventional footwear.

As displayed in Figure 25, this would suggest that there is a scale of similarity away from barefoot upon which types of minimalist footwear and conventional footwear would sit. The most minimalist footwear, offering the least interference through the most flexible structure and thinnest sole would be closest to barefoot on this continuum whilst as the degree of structure, aesthetic design and/or protection is increased they would move away from the barefoot end of the spectrum. It seems however that whilst footwear companies should strive to deliver a shoe which gets as close to barefoot as possible, the shear nature of placing a protective surface between the foot and the ground will always result in a difference between walking barefoot and walking in any type of footwear. It can be speculated that this is due to an inherent cutaneous afferent interference, both on the dorsal or plantar portion of the foot, via the material encasing the foot or through a psychological component of perceived protection in comparison to a bare foot. Nonetheless it is apparent that walking in minimalist footwear and walking barefoot will never be entirely comparable and minimalist footwear cannot be marketed as replicating barefoot walking conditions only offering the most viable alternative. This could in effect be the best option however, offering similar kinematics to walking barefoot whilst offering a small degree of protection against abrasions.

New technologies such as 3D printing could be a great way for the footwear industry to improve upon their desire to provide the optimal shoe and/or get as close to barefoot as possible. The potential to match footwear to foot shape using this technology could maximise fit and comfort and also have potentially wider benefits. The importance of comfort has been proposed recently by Nigg et al (2015) who explains how being comfortable, through in this case selecting the most comfortable shoe, allows runners to adopt their preferred movement path and reduce injury risk. This proposal could also be applicable to the findings of this study whereby offering a shoe which was more flexible and in some cases had better fit allowed participants to feel more comfortable, revert to their preferred movement path and experience a reduction in joint pain symptoms and improved balance confidence. Particularly for older adults suffering from foot problems leading to abnormal foot shapes and types, 3D printing to match a footwear to this shape could be of particular benefit.

Building upon the notion of the optimal footwear, there are numerous factors which would influence this and in effect should be determined on a case by case basis. The time of year and consequently the weather has a large bearing on footwear choices and what would be considered the best option for an individual. For example, during winter a minimalist shoe which is as close to walking barefoot as possible would likely not be considered a feasible option for most individuals given the cold climate and probable wet and slippery conditions underfoot. In this case a shoe offering greater warmth, water resistance and greater sole grip would be the optimal choice. Conversely in the middle of summer with a hot climate, high humidity and reduced risk of treacherous walking conditions the reverse would be true and most individuals would likely be happy to walk in a shoe more akin to walking barefooted.

Similarly previous footwear habits would play a part in the successfulness of the adaptation to minimalist footwear. A 65 year old lady who has spent the last 40 years wearing a heeled dress shoe may find the transition straight to a minimalist shoe of great similarity to barefoot uncomfortable due to the drastic change in foot position and the inherent changes to leg and body alignment which may have occurred as a result of her past footwear history. Previous research has explored this effect of high heels on the musculo-skeletal system and demonstrated how prolonged wearing of high heels can result in the shortening of the muscle fascicles within the triceps surae as well as a stiffening of the Achilles tendon leading to a reduction in the ankles active range of motion (Csapo, Maganaris et al. 2010). This could considerably impair the effective transition to a low/non heeled shoe such as a minimalist footwear. In this case starting with a minimalist shoe closer to conventional footwear would allow for a smoother transition to natural posture and motion through a gradual change in musculoskeletal alignment. As such the offering of a range of minimalist shoes along the continuum, as displayed in Figure 25, is of benefit in order to satisfy the various constraints that arise when choosing footwear. This could be perceived as a limitation to the study presented in Chapter 7 as only one type of minimalist shoes was able to be provided to participants to wear for the 4 month intervention period. Due to the nature of daily life, the suitability of one type of footwear for all occasions and conditions is unlikely and this suggestion was supported by participant feedback. During tasks or conditions where the participant felt it unsuitable for them to wear the minimalist shoes provided to them, such as particularly wet weather or rough off-road terrain, they substituted them for their conventional footwear conditions. It is therefore feasible that through greater use of minimalist footwear in more challenging conditions could have evoked greater changes and thus it is recommended for future research to provide more than one option of minimalist footwear encompassing a range of the continuum in order to improve real life applicability and increase recruitment and adherence.

This thesis has shown that functional and static balance improvements can result from increasing the activation of intrinsic foot muscles. During repetitive activation of the foot muscles, as per a training protocol, not only strength can be improved but also the control and feeling of the toes can be enhanced or recovered. This aspect is concomitant to muscle activation and may be as important for stability and confidence as the strength improvement itself and may have contributed to the balance improvements reported the training study in Chapter 6 and the footwear intervention study in Chapter 7. This confirmed the important role that our feet and toes play in providing feedback regarding our body's position in relation to the ground alongside providing our base of support. Consequently ensuring the foot remains an active, sensitive and strong structure can have positive consequences on stability and training interventions like this one should be implemented.

This thesis also showed promise for the use of minimalist footwear in daily life as a method to increase foot muscle activation as well as functional balance ability and balance confidence. As previously stated increased muscle activation can have benefits to both improved strength but also feeling and control. Whilst we saw a small improvement in hallux plantar flexor strength following the use of the minimalist footwear there were also reports from participants of improved feeling and movement across the foot. This occurred within only a 4 month intervention period and thus it would be particularly interesting to see the long term effects in this population with continued use of minimalist footwear in foot mobility and feeling and also overall balance and gait parameters.

Alongside this proposed strength benefit it was highlighted however that through wearing this type of shoe reductions in painful symptoms at the foot and more proximal joints such as the hip were reported. With this in mind to build upon this work more participants suffering from painful foot and joint conditions should be recruited to determine if they may experience some relief through minimalist footwear. It can also be speculated that long term use of minimalist footwear throughout the lifespan and prior to older age could be beneficial in reducing the progression of arthritic conditions and the prevalence of common foot problems associated with conventional footwear. Thus research into the use of minimalist footwear throughout the lifespan may also be warranted.

Ultimately this thesis has demonstrated that there is promise in the lifestyle use of minimalist footwear for a healthy older age population in terms of; i) the potential to improve muscle strength, feeling and control at the foot and lower leg muscle level; ii) an improvement in balance confidence due to an increased feel of the ground potentially leading to increased physical activity; iii) a reduction in the foot and joint pain experienced. These are areas which warrant exploration in future work across various populations to enhance our understanding of whether feet and footwear are friends or foes.

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