BIOMECHANICS OF ASSISTED LOCOMOTION IN ELDERLY OSTEOARTHRITIS PATIENTS

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Finally, but by no means least, a very big thank you goes to my mum, without whom this would never have been possible. I dedicate this thesis to her

ABSTRACT

Osteoarthritis is the most widespread musculoskeletal disease worldwide among the elderly. It causes joint pain that can affect locomotion and reduce mobility. For osteoarthritis patients, maintaining walking ability is considered the most beneficial way to preserve their quality of life. Walking sticks are widely used by elderly adults and have been shown to have a supportive role on locomotion. In this thesis I carried out four experiments: The first study investigated how footwear affects the locomotion of elderly patients suffering from this disease. In the second chapter the gait of elderly walking stick users was analysed in conjunction with their responses to a questionnaire with a view to understanding the causes and context of walking stick use in their everyday environments. My findings demonstrated that the majority of participants experienced greater pain after prolonged use of their walking stick. In the last two experiments I investigated how the use of a walking stick combined with aspects of the individual's locomotor environment (e.g. indoor and outdoor, level and sloped surfaces) to influence gait. Overall, osteoarthritis, advanced age and challenging locomotor environments can influence their quality of life and the risk of falling.

Keywords: Locomotion, Osteoarthritis, Walking sticks, falls

CHAPTER 1. GENERAL INTRODUCTION

BIOMECHANICS OF ASSISTED LOCOMOTION IN ELDERLY OSTEOARTHRITIS PATIENTS

1. General introduction

1.1. Human locomotion

1.1.1. *The gait cycle*

Human locomotion is an essential activity required in order to move and explore the environment in which we live and can be defined as the movement from one place to another expressed in several different forms: walking and running, swimming, or even riding a bike (Morecki et al., 1997). Humans can walk independently by the age of 11 months (Sutherland et al., 1988). This activity is generated by the pattern of rhythmic alternating movements of the lower limbs creating forward body movement and is described as human gait (Shulz et al., 2005). The notions, however, of human walking and human gait are not identical. Whittle (2007) makes a distinction between these two terms suggesting that human walking or running is a process requiring the use of two legs which creates a propulsion similar to an inverted pendulum over the foot contacting the ground (Whittle, 2007, p. 48), whereas gait is characterized as the style referring to the manner of the walking, the pattern description and not the process itself. Walking is the process of moving forward, a gait pattern which involves the combined alternation of the two lower limbs and the succession of double and single support phases so that the body remains permanently in contact with the ground and is effected through the integration of visual, vestibular and proprioceptive input controlling and coordinating the musculoskeletal system of the body (Hausdorff, 2007, p. 556). According to Perry (1992) there are three locomotor functions i) propulsion, ii) balance during the shock absorption and iii) energy conservation and if at least one of these functions is deficient then gait is impaired.

In forward walking, the sequence between the moments when one foot strikes the ground until the moment when the same foot contacts the ground again is defined as a gait cycle or simply as a stride (Whittle, 2002). The analytical examination of the gait cycle is the main form of description of normal and pathological gait. The gait cycle comprises two phases: the stance phase and the swing phase. The former is initiated with the one foot contacting the ground with a heel strike and terminates with toe-off when the foot is lifted from the ground, corresponding commonly to 0 to 60% of the entire gait cycle duration (Whittle, 2002; Richards, 2008). Subsequently, the swing phase of this foot is initiated and the foot moves forward to contact the ground again in a double support phase when both feet are on the ground, corresponding to the second part of the gait cycle (60%-100%) represents approximately the 40% of the entire duration (Murray et al., 1964; Sutherland et al., 1994; Richards, 2008).

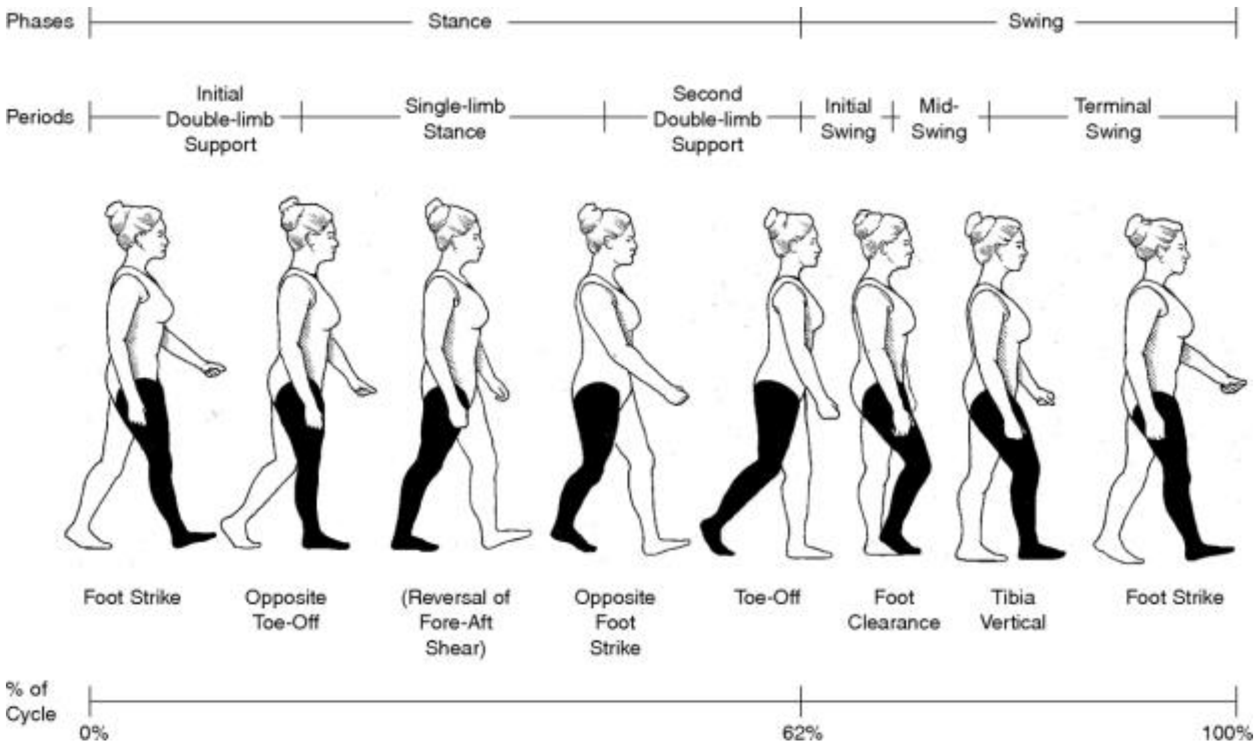

Figure 1. Illustration of normal gait cycle by the right leg (black) describing the detailed stance and swing phases (Sutherland et al., 1994)

The stance phase starts with the initial contact (0% to 2% of the gait cycle): the ankle is in moderate plantar flexion (Figure 1) and a vertical force is applied at the heel during the loading response (0% to 10% of the gait cycle) which passes through the knee and the hip. This force vector tends to extend the ankle and flex the hip as shown in Figure 2. The heel strike is then followed by the foot being placed in a flat position. This involves a short phase where plantar flexion is restricted by the action of the levator muscles (tibialis anterior and extensor) and mid-stance (10% to 30% of the gait cycle) in which the foot is flat. The stance phase ends with the opposite foot strike (30% to 50% of the gait cycle) and toe-off when the heel is lifted (50% to 60% of the gait cycle). Muscles controlling the end of the stance phase prepare the lower limbs for the swing phase. Push off is effected with knee flexion, which is simply related to the inertia of the tibial segment, and hip flexion gives the essential propulsion to switch the step (Murray et al., 1964; Winter, 1983; Sutherland et al., 1994; Richards, 2008).

The second phase of the gait cycle beginning at toe-off is called the swing phase. It is divided into three phases: First the initial swing starting at approximately 60% to 73% of the gait cycle when the foot leaves the ground; hip flexion allows the progression of the leg and knee flexion is done automatically the inertia of the leg with respect to the thigh moving forward under the activation of the hip flexor muscles (iliopsoas, adductor longus, satrorius and gracilis). The next phase is a sequence between the accelerating phase and the decelerating phases corresponding approximately to 73% to 87% of the gait cycle (the variation around the mean can diverge) and involves foot clearance to the ground where the ankle is in a neutral position due to the tibialis anterior muscle and knee in maximum flexion, whilst the hip is also in flexion. Finally, the terminal swing phase (87% to 100% of the gait cycle) is when the hip flexion is in deceleration (Murray et al., 1964; Winter, 1983; Sutherland et al., 1994; Richards, 2008).

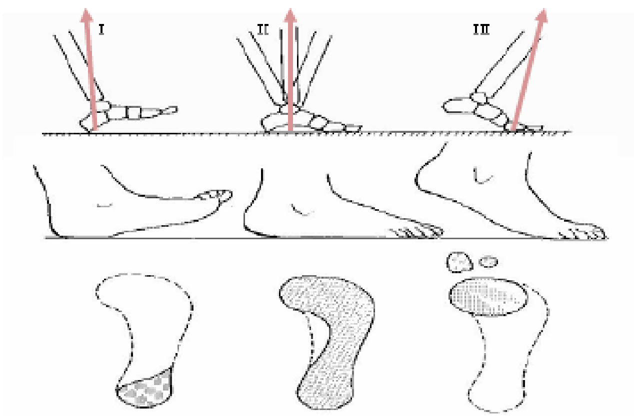

Figure 2. Representation of the force vectors and pressure distribution during heel strike, foot-flat and toe-off phases (from Plas et al., 1989)

1.1.2 Gait analysis

Gait analysis in a clinical environment is a quantitative method of data collection analysing walking patterns and seeking to elucidate the aetiology of gait disorders (Davis et al., 2002). It is through this process that the variables defining walking abnormalities can be identified. Clinical gait analysis performed in the lab uses accurate measurement tools such as the Vicon system with infrared cameras and applied markers on the body segments, giving accurate results on the characterisation of walking pattern. Nevertheless, the significance of observational gait analysis outside a lab setting should not be underestimated. Additional ecological data based on visual images (ie video gait analysis in a real environment) can provide a better understanding of each individual's pathological condition (Davis et al., 2002). Kinematics analyse the movement mechanics describing the position of points in space (spatial parameters) and time (temporal parameters) (Richards et al., 2013).

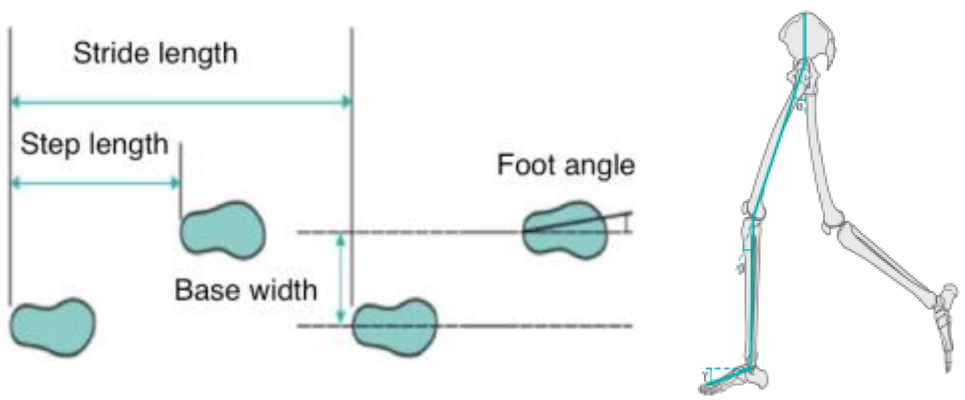

Figure 3. Kinematics gait analysis: Definition of gait spatiotemporal parameters (left) and joint angle measurement (right) (Richards et al., 2013)

As shown in Figure 3, step length is the distance between two successive heel strikes and stride length is the distance between two heel strikes of the same foot or the sum of left and right step lengths. Respectively, step time is the time span between two successive heel strikes and stride time is the time between two heel strikes of the same foot or the duration of one gait cycle. Base or step width is the distance of the medial - lateral separation of the centre of the two heels. Cadence is the number of steps per minute. Gait speed (m/s) can subsequently be calculated as the product of cadence (steps/min) and the step or stride length (m) divided by the number of seconds in one minute. Foot angles measure the orientation and foot motion in the line of progression over ground. Joint angles can be defined as the line between the proximal and distal segments of a joint. Figure 3B displays the joint angles of the lower limbs (α,β,γ). The ankle joint angle (plantar-dorsiflexion) is the calculated angle between the foot and tibia (Figure 4). The knee joint angle (flexion/extension) is the angle shaped between the segments of tibia and femur (Figure 5) and the hip angle (flexion/extension) between the segments of pelvis and femur (Figure 6) (Richards et al., 2013).

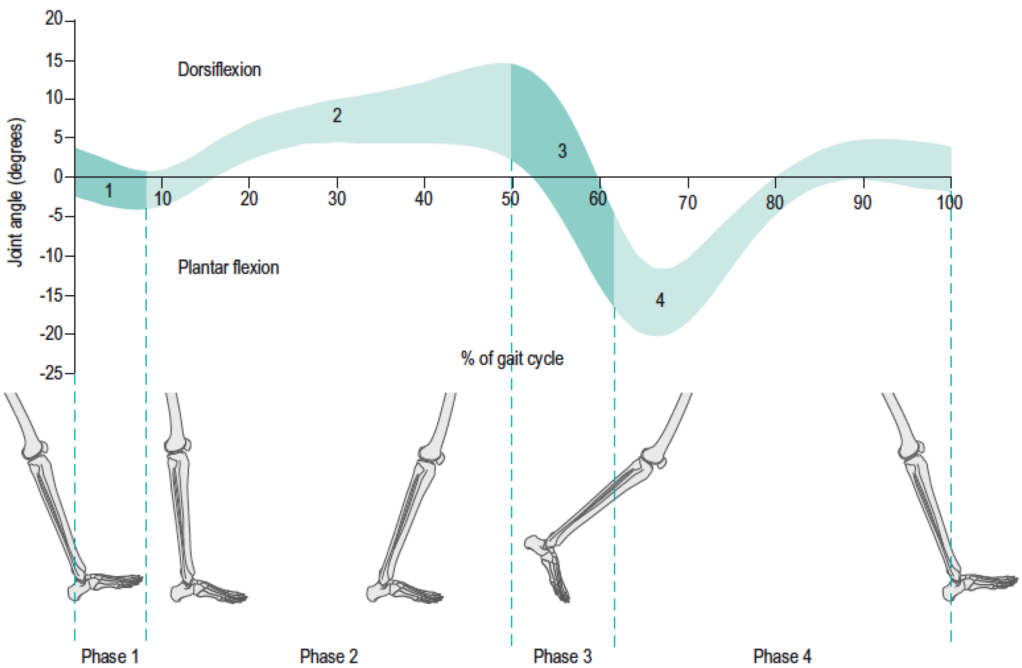

Figure 4. Plantar dorsiflexion of ankle joint angle during the gait cycle (Richards et al., 2013)

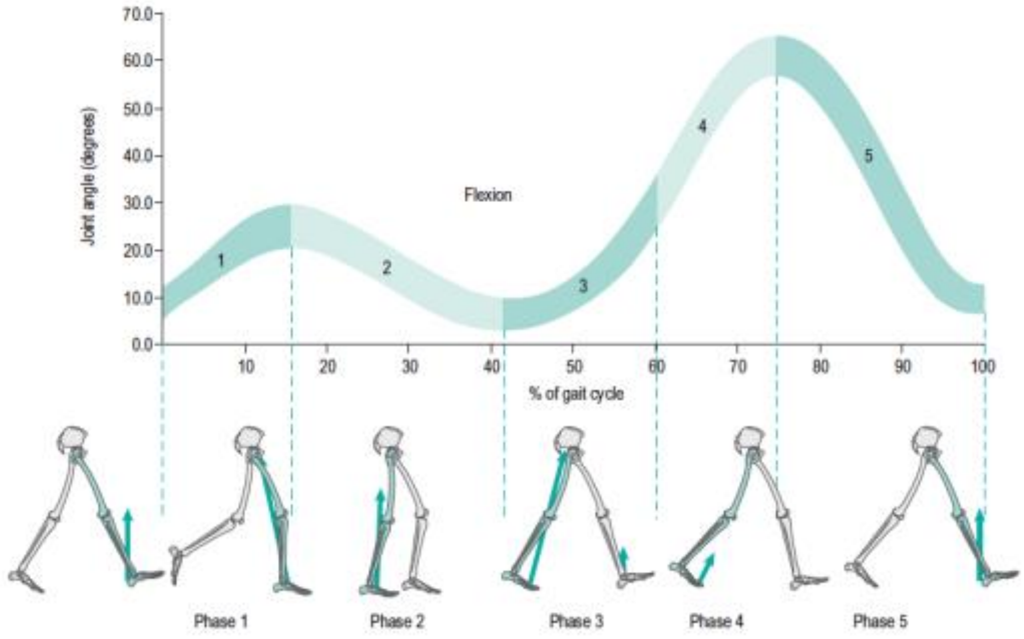

Figure 5. Knee flexion extension joint angle during the gait cycle (Richards et al., 2013)

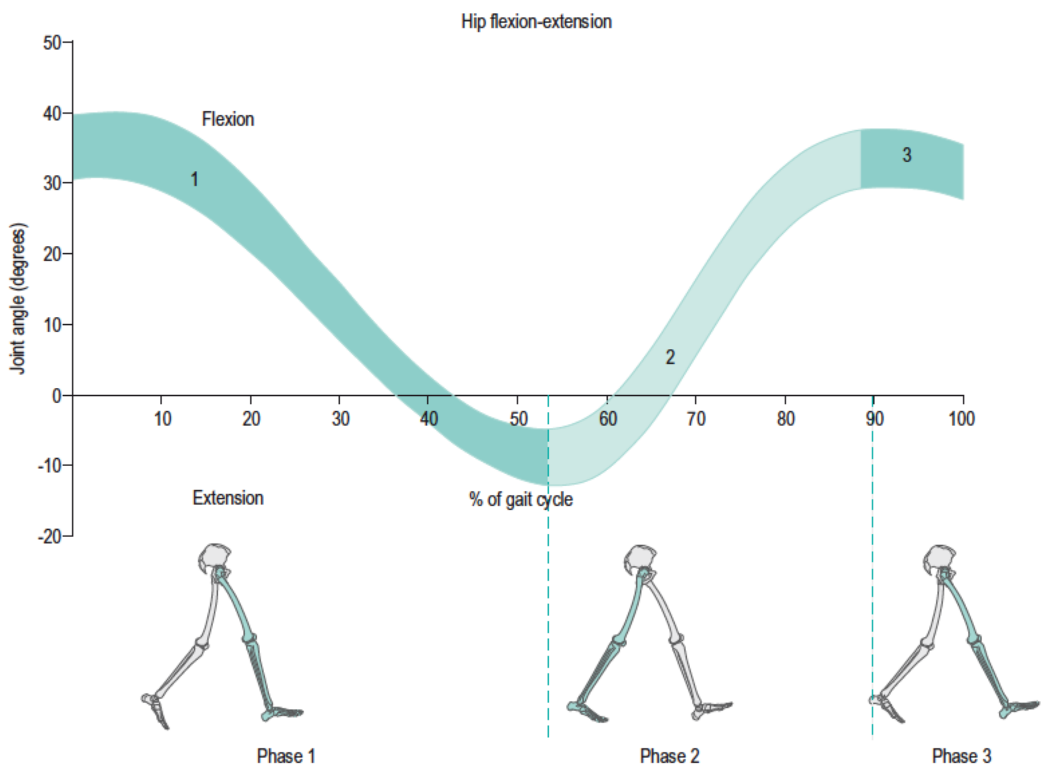

Figure 6. Hip flexion extension joint angle during the gait cycle (Richards et al., 2013)

In the literature gait analysis is mainly focused on the lower limbs as they are the principal elements of walking movement. However upper limb kinematics specially the head and the trunk shouldn't be neglected from gait analysis as they can give indications of balance and gait stability during walking activities (Cappozzo et al., 1981; Carlson et al., 1981; Breniere et al., 1998). The trunk is a large segment positioned above the feet, with many muscles and articulations, representing approximately 60% of the body mass that participates in the majority of motor activities (Carlson et al., 1981; Breniere et al., 1998; Li et al., 2006). The analysis of trunk motion can indicate pathological gait (Krebs et al., 1992; Moe-Nilssen et al., 2005; Houck et al., 2006; Engsberg et al., 2009). According to Gracovetsky (1988) the trunk motion regulates gait. In fact, trunk flexion can affect the mechanical energy flow in the lower extremities during gait and it can therefore reduce walking efficiency (Takeda et al., 2016). Forward flexion of the trunk can be a destabilising parameter because it shifts the centre of gravity, increasing instability and the risk of falling (Krebs et al., 1992). The efficient human locomotion involves the smooth forward movement of the centre of mass by the conversion of mechanical energy into kinetic energy during the displacement of the centre of mass (Takeda et al., 2016; Cappozzo et al., 1976; Perry et al., 1974). Trunk-flexed gait is characterized by a forward leaned body position, increased knee flexion, ankle dorsiflexion and hip flexion during the stance phase (Saha et al., 2008). This gait pattern with the mentioned kinematic alterations requires higher mechanical energy than normal gait and it can consequently be less stable (Takeda et al., 2016).

Kinematics is a branch of mechanics that estimates the spatial position and/or its derivatives when one or more boundary conditions are known. Although kinematics quantify motion during gait they do not take into account the causes of motion. Instead, they analyse only the position, its derivates (speed, acceleration) and the angles of movement. Kinetics can explain the system's state, the relationship between the motion of points and their causes, underlying mechanisms such as forces, and moments that create motion. Kinematics can be collected with motion capture systems such as the Vicon system and kinetics with force plates (Richards et al., 2013). A force plate is used to collect force data applied on its surface when the foot contacts the plate (Figure 7).

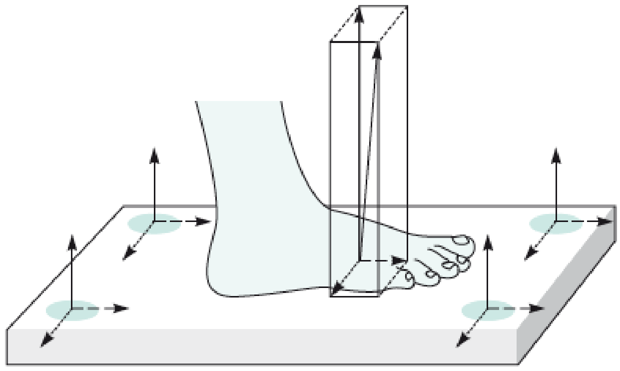

Figure 7. Kinetics gait analysis: Ground reaction forces of the foot strike on the instrumented force plate (Richards et al., 2013).

Force is a vector with magnitude and direction that, when applied by the body on the ground, can create propulsion or braking during gait (Richards et al., 2013). Kinetic variables such as ground reaction forces can characterize normal and pathological gait as the cause of that particular gait (Winter, 1991; Perry, 1992). In the analysis of a vertical ground reaction force (Fz), there are some important points to be taken into consideration. The vertical force Fz is characterized by a double peak, starting with heel contact with the ground followed by the first peak which corresponds to maximal weight acceptance. At this point, when the knee is in flexion, the curve representing the force drops as the weight is unloaded during mid stance until the propulsion during push off is produced by the plantarflexor muscles, showing the increase of upward speed. Finally, the applied force is reduced again until the foot is lifted at toe off and weight is transferred to the opposite leg. The shape of vertical ground reaction force is illustrated in Figure 8.

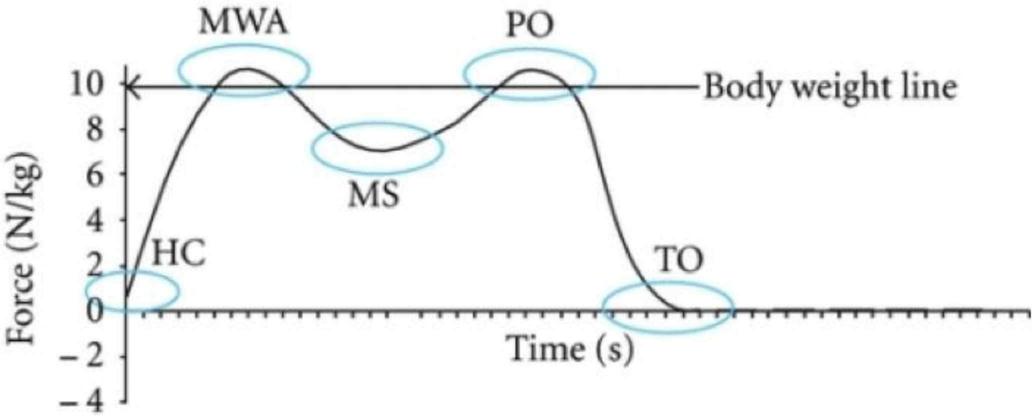

Figure 8. Profile of the shape of vertical ground reaction force during the stance phase of a normal gait pattern. Heel contact (HC), maximal weight acceptance (MWA), mid stance (MS), Push-off (PO), and Toe off (TO). (Bouffard et al., 2011)

The analysis of ground reaction forces provided important data describing the causes of movement and gait abnormalities. However, kinetic data are generally collected in a lab, under strict experimental conditions. Since the participants have to step on the centre of the force plate in order to obtain accurate data, they might adapt their step length in order to precisely place their dominant foot on the plate (Challis, 2001), and thus modify their natural gait. This adaptation can provide accurate results for the clinical research (Medved, 2001) but it is an error that can interfere with the ecological studies. Therefore, I will use it in our first study in order to have a deep insight in the biomechanical differences between barefoot and shod walking. However, in the next chapters we will investigate natural locomotion, in outdoor or sloped conditions. Thus, I will focus on the analysis of kinematics without integrating a force plate in the experimental setting in order to obtain the natural locomotor pattern and the age related differences.

1.1.3 Gait changes in the elderly

The literature suggests that as we get older we tend to walk slower, with shorter steps with longer stance phases and we increase the time that both feet contact the ground in order to increase our stability (Gabell & Nayak, 1984; Imms & Edholm, 1981; Winter, Patla, Frank, & Walt, 1990). The human ageing process is associated with a number of neuromuscular changes, among which are those of the sensory-motor system affecting the control of gait and posture and can be the reason for the deterioration of balance capacity (Horak et al 1989 Woollacott. 2000). Previous studies have found that the locomotor system is affected by a decrease in visual acuity (clarity of vision) (Bohannon et al. 1984), a reduction in range of joint motion and lower proprioception (Skinner et al.1984, Shaffer & Harrison 2007). Proprioception can be defined as the somatosensory information providing the sense of body position, spatial orientation and movement (Sarlega et al., 2009). It should also be pointed out that the vestibular system is also affected by ageing (Woollacott & Tang 1997). In general, the literature shows that following proprioceptive and vestibular impairment, the elderly develop a greater reliance on vision for postural control (Lord & Menz 200). Furthermore, the ability to integrate multi sensory signals in the brain is reduced, thus weakening the capacity to select information (Woollacott et al. 1986) and the decrease in nerve conduction speed results in a delay in feedback (ie visual feedback (Woollacott & Tang 1997) and in postural reflexes (Dorfman & Bosley 1979). These degenerative changes may increase instability and amplitudes of postural sway, consequently altering gait equilibrium in a number of ways (Horak et al., 1989, Alexander et al. 1992).

Furthermore, as we age, muscle mass gradually decreases with the degeneration of certain fibers (mostly Type II fast twitch fibers generating strength, speed and resistance to fatigue), creating the phenomenon called 'sarcopenia' which is partly due to a general degeneration of the nerve tissue (Doherty & Brown 1993). In spite of the fact that this decrease of fibers could theoretically be compensated with physical training (Asmussen 1973; Bouisset Matton & 1995), it is hard to establish a link between lack of physical activity and muscle atrophy. Thus, the major consequence of the above phenomenon is the reduction of intrinsic strength, due to the combined effects of the following modifications in: (i) muscle architecture, (ii) mechanical / elastic properties of tendons, (iii) conductivity of neurons (weaker agonist muscles compared to antagonists), and (iv) the intrinsic elemental force developed by each fiber. More recently, it has been demonstrated that the changes of tendon mechanical properties of plantar flexor muscles have contributed to the reduction of muscle strength, caused mainly by the increase of stiffness of tendons (Magnusson et al 2008; Narici et al. 2008). Several studies have reported a decrease of maximum plantar flexion angle (Vandervoort & McComas 1986; Thelen et al. 1996; Ferri et al. 2003; Kubo et al. 2008), a reduction in the ankle joint range of motion (Kubo et al., 2007), and a decrease in walking speed due to the shortening of muscle in the elderly (Ochala et al 2004).

The cause of gait impairments and the reduction of walking stability in the elderly is usually multifactorial. The combined factors include sensory deficits caused in vision and the vestibular system, degenerative diseases and the psychological effect of anxiety and fear of falling (Jahn et al., 2010). It is, therefore, crucial to investigate these factors, as gait impairments and instability can increase the risk of falling, ultimately leading to a reduction of quality of life and even mortality (Jahn et al., 2010).

1.1.4 Falls in the elderly

Due to the demographic growth of the population and the continuing increase in life expectancy, it is essential to investigate the issue of falls in the elderly which have far-reaching consequences at an economic, functional (hip fracture, death), and psychological (fear of falling, social isolation, loss of independence) level (Albertsen & Temprado, 2011). Approximately one third of adults over 65 years experience a fall at least once a year (Tinetti, 2014); about 16% of emergency incidents in the elderly are caused by a fall (Tinetti, 2014); in addition, the vast majority (90%) of domestic accidents in the elderly are cause by falls (Tavernier -Vidal et al, 2014). A study by Rubenstein suggests that women fall 2 to 4 times more often than men, but this trend decreases as age advances, to arrive at almost at the same proportion of males to females after the age of 85. Furthermore, a person who has experienced a first fall runs a 20 times greater risk of falling again. Finally, the risk of falling is three times higher among the elderly living in a nursing home (Rubenstein et al, 1994).

It has been demonstrated that age, certain types of medication (including antidepressants and neuroleptics), gait instability, decrease in vision, fear of falling and diseases such as dementia and osteoarthritis are among the most important risk factors for falls (Kallin et al, 2002; Tavernier -Vidal et al, 2014). Environmental factors which can increase the risk of a fall include cluttered rooms with obstacles, slippery or uneven surfaces, insufficient lighting conditions, and unsuitable footwear

(Tavernier -Vidal et al, 2014). Moreover, it is important to stress the psychosocial consequences of falls: although 85 % of falls do not cause physical injury, the psychosocial consequences can be serious as they can restrict the number of daily activities (Grisso et al, 1992; Pfitzenmeyer et al, 2001), reduce confidence, independence, and increase the fear of falling (Tinetti et al, 1994; Pfitzenmeyer et al, 2001). The risk of falling can increase also because of extrinsic factors such as reduced lighting, slippery or uneven surfaces, cluttered rooms, footwear (Lord et al., 2000) or inappropriate mobility aids (Dean et al., 1993). This thesis will investigate both intrinsic and extrinsic factors increasing the risk of falling in the elderly starting from osteoarthritis the most prevalent musculoskeletal disease in the elderly population.

1.2. Overview of osteoarthritis and walking stick use

1.2.1 Osteoarthritis in the elderly

Arthritis is a common condition that causes pain, inflammation, functional limitation in a joint and it is the most widespread musculoskeletal chronic disease worldwide among the elderly (Felson et al., 1995; Brooks, 2006; Turkiewicz et al., 2014). According to the NHS, in the UK approximately 10 million people suffer from arthritis and the vast majority, approximately 8 million people, are affected by osteoarthritis (NHS, 2016), a degenerative disease usually afflicting middle-aged adults who are 50 years old or older, with individuals having a family history of the condition and women being more prone (MMWR, 2013). Initially it affects the smooth cartilage lining of the joint, restricting the range of movement, causing pain and rigidity. Swelling and bone deformation may ensue as the cartilage lining becomes stiffer and weakens and there is an overload of tendons and ligaments which are more solicited (Minor & Kay, 2009). Severe damage of cartilage can cause bone rubbing against other bone, leading to modifications of bone position and joint shape (Minor & Kay, 2009). Furthermore, the sub-chondral bone of an osteoarthritic joint thickens, and may result in the appearance of osteophytes and sub-chondral cysts (Figure 9), thus modifying the structural integrity of the affected joint (Minor & Kay, 2009). Joints most frequently affected are those located in the hands, spine, knee and hip (Minor & Kay, 2009). As osteoarthritis gradually evolves with age, pain in the affected joints becomes more severe (Conaghan et al., 2009). Consequently, osteoarthritis is undoubtedly associated with ageing and increased mechanical stress on the joints (Berenbaum et al., 2013), affecting everyday life activities including automatic movements such as walking, and can have an intensifying effect on age-related gait disturbances (Whittle, 2007). The severity of the disease differs and can be rated from 0 (none present) to 4 (severe) (Arden et al., 2006), though it may be quite misleading to evaluate data from this scale as the pain threshold in each individual varies immensely. From a clinical perspective, osteoarthritis can be detected radiographically and it is characterised by degenerated joints (Altman et al., 1986). Initially, osteoarthritis pain appears only during or after practising a physical activity with high intensity and it becomes worse over time. Gradually, the pain becomes constant and more severe over time affecting a bigger range of lower intensity activities such as walking or grasping objects (Fautrel et al., 2005). The development of the disease, including the pain and loss of function for example, lead osteoarthritis patients to rely on others in order to perform daily living activities (Gooberman-Hill et al, 2009). In order to compensate for osteoarthritis pain, elderly patients tend to alter their gait pattern (Chen et al., 2008). These changes may decrease general movement and thus independence (Whittle, 2007) and can increase the risk of falling (Sturnieks et al., 2004; Levinger et al., 2011). According to NICE (2014), osteoarthritis can be treated by joint surgery or by pharmaceutical use of oral analgesics and topical non-steroidal anti-inflammatory medication, although maintaining walking ability and being active is considered the most beneficial and effective way to preserve patients' quality of life. In addition, maintaining the ability to walk places no financial burden on either patients or the health care system.

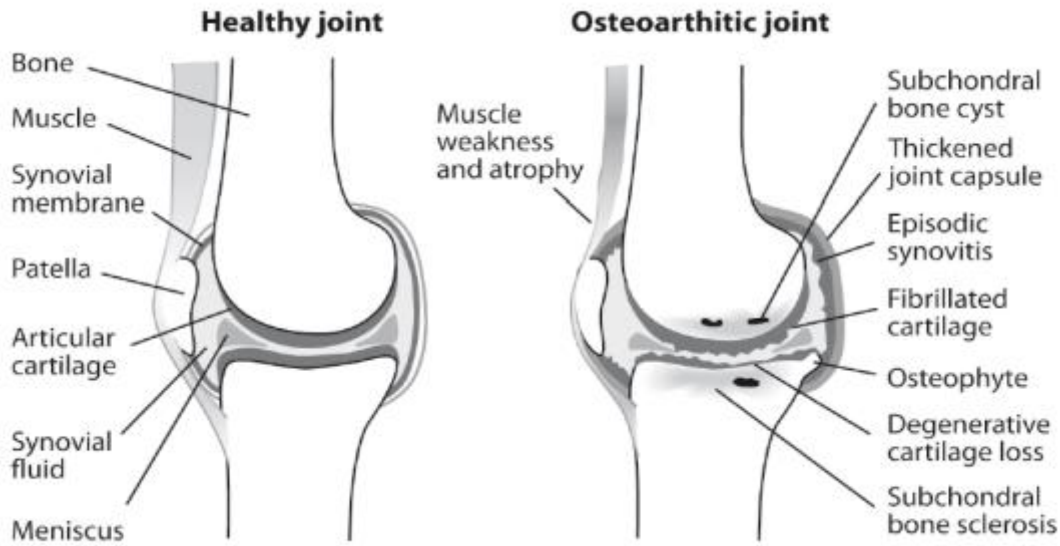

Figure 9. *Illustration of a healthy (left) and an osteoarthritic joint (right) (Arthritis Research UK).*

According to Ickinger & Tikly (2014), osteoarthritis falls into two categories, depending on the causal factors of the disease. Primary osteoarthritis can be attributed to genetics or normal ageing and secondary osteoarthritis can be a result of previous injuries or other pathological conditions in the bones, joints or metabolic disorders. Previous literature has suggested that having a previous joint injury (Gelber et al., 2000), heavy weight loading on the joints (Conaghan, 2002), other metabolic disorders or obesity (McAlindon et al., 1999) can potentially cause osteoarthritis, although the human ageing process is considered to be the greatest risk factor of the disease (Minor & Kay, 2009).

Osteoarthritis has been classified among the five most important impairments in the elderly population (Murray & Lopez, 1997) and has an important economic impact on our societies. In the UK, as the elderly population and life expectancy increase, it is a crucial challenge for the healthcare services to support osteoarthritis patients, improve their quality of life and reduce the prevalence of the disease (Statistics, 2011). According to the NHS, approximately 6.5 million patients suffer from hip and knee osteoarthritis. In England, approximately one in five adults (18.2%) aged over 45 has osteoarthritis of the knee and this form of the disease accounts for the majority (98%) of knee replacement surgery. Respectively, one in nine adults (10.9%) over 45 has been affected by osteoarthritis of the hip joint. In fact, it is estimated that the percentage of individuals with mild impairments caused by osteoarthritis is higher as some of those affected do not seek or follow medical advice (Arthritis Research UK). The knee and the hip play a key role in human locomotion and the gait cycle specifically - damage caused by osteoarthritis on these joints is very likely to affect gait.

Chen et al. (2012), investigating the global financial impact of osteoarthritis, divide spending into three categories: firstly, direct costs pertaining to hospitalization expenses, pharmacological treatment and surgery; secondly, the indirect cost arising from the effects of the disease and the caused pain or impairments to the patients and can be defined as lack of productivity or absence from work, the risk of falls and fatal accidents; thirdly, indirect costs include the increased budget required for the compensatory benefits provided to disabled patients. It should, however, be pointed out that the impact of osteoarthritis is not only financial. Pain, reduced independence and deterioration of quality of life are the intangible costs and it is difficult to calculate their exact cost to the health and social care system (Xie et al., 2008).

In view of the above, it is vital to investigate gait in osteoarthritis patients in order to provide a better understanding of the biomechanical factors responsible for making the disease more acute and painful for patients (Radin et al., 1972). The aim of this thesis is, therefore, to fill the gap in gait biomechanics of elderly osteoarthritis patients using a walking stick and thus to contribute to a better understanding of the underlying mechanisms leading to falls, reduction of walking ability and, consequently, quality of life.

1.2.2 Gait changes in osteoarthritis patients

The literature suggests that osteoarthritis can extend the effects of ageing in gait in spatiotemporal parameters causing further deteriorations in the gait ability (Mandeville et al., 2008; Astephen et al., 2008; Ko et al., 2011). As a degenerative disease, osteoarthritis can be the origin of a pathological gait, an abnormal walking pattern resulting from muscle weakness and pain in the joints (Malanga et al., 1998). Gait impairments can be attributed to neurological or musculoskeletal conditions and deficient motor control can be displayed through alterations in gait parameters of osteoarthritis patients. Arthritic joints present a reduced range of motion which restrict normal movement and increase the duration of the gait cycle (Hurwitz et al., 2000). Walking speed is reduced and the duration of the stance phase, especially of the affected limb, increases. In fact, the body position as a whole is affected by the abnormal gait pattern. The hip extensors are over-solicited and the trunk is flexed in a forward direction. Therefore, the centre of gravity can be shifted as the osteoarthritis patient walks with an increased forward leaning position (Neumann, 2013). The mechanical stresses on the arthritic joints are higher. The arthritic hip presents excessive flexion to compensate for excessive knee flexion and ankle dorsiflexion. The arthritic knee presents decreased extension angle, affecting heel strike which is restricted and causing, therefore, a reduction in step and stride length. The restricted heel strike in turn causes a reduction of floor clearance during the swing phase of gait. The above alterations increase instability during walking (Chen et al., 2008).

However, research has not concluded whether these gait alterations are associated only with the severity of osteoarthritis or with the restricted range of movement, pain and general weakness regardless of the fact that these are highly linked (Hurwitz et al., 2000; Mc Gibbon et al., 2002). Gök et al. (2002) suggest that it is a multifactorial effect on gait pattern affecting other joints of the lower limbs and the lower back. Overall, osteoarthritis patients develop a walking pattern characterized by decreased walking speed, decreased cadence, shorter step and stride length, and shorter stance phase of the affected limb (Gök et al., 2002). Therefore, it would appear very important for osteoarthritis patients to unload the affected joints from mechanical stress, to achieve better balance and support during gait in order to maintain a reasonable level of physical activity and reduce their risk of falling. This thesis aims to investigate the interaction of the elderly with the environment, to understand the multifactorial decline of physical function during the ageing process and to focus on the factors that can improve or influence gait. For the elderly, two critical contact points with the ground can be identified, shoes and walking sticks, as they not only regulate the contact of the body with the ground but are also two important sources of sensorimotor input.

1.2.3 Walking sticks compensating osteoarthritis

It is estimated that 24% of the elderly population over the age of 65 make use of walking sticks, in order to feel safer and more stable, (Resnik et al., 2009). Previous studies have shown that 40–76% of the elderly osteoarthritis patients use a walking stick for better support during gait and to alleviate osteoarthritis pain (Van der Esch et al., 2003; Shrier et al., 2006). Clinical findings suggest that walking sticks are beneficial in improving movement and stability in everyday activities (Blount, 1956; Murray et al., 1969; Bennett et al., 1979). In fact, using a walking stick can remove weight from the affected joints and assist in maintaining stability during gait (Mulley, 1988). In their study, Mulley (1988) suggested that on average approximately 13.5 % of the body weight of an osteoarthritis patient is applied on the walking stick. This additional point of contact with the ground can regulate mass distribution and better adjust weight loading, leading to a more balanced gait pattern (Whittle, 1998). By reducing the load on the affected limb (eg knee, hip) with the use of a walking stick, osteoarthritis pain can be countered, allowing the individual to have improved stability (Figure 10) and motor control of the lower limbs during gait (Bateni & Maki, 2005). Furthermore, clinical findings have shown that using a walking stick can have a positive effect on locomotion after rehabilitation of an injury (Engel et al., 1983).

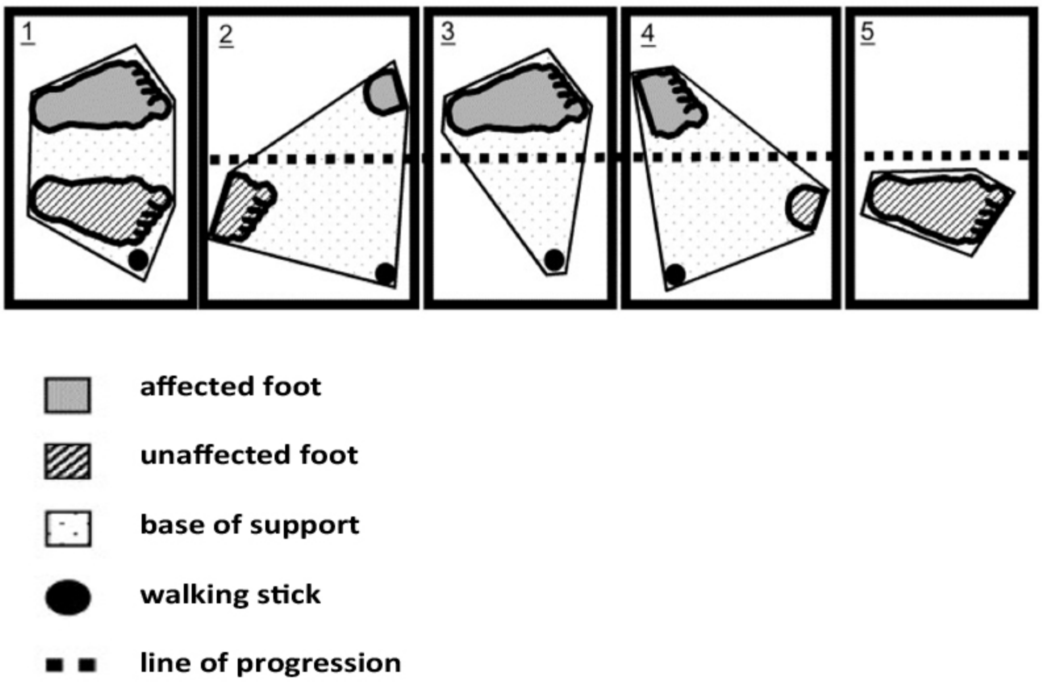

Figure 10. *Illustration depicting how the use of a walking stick can increase stability by providing a higher base of support while standing and walking. Panels 1 to 5 describe contralateral walking stick use: (1) standing posture supported by a walking stick; (2) walking stick in deceleration; (3) affected foot and walking stick strike (4) walking stick in propulsion; (5) non-affected foot strike and walking stick in forward displacement without ground contact (Bateni & Maki, 2005).*

The centre of mass is the central point of the totality of gravity force applied over the body. The regulation of its displacement over the base of support can lead to a stable gait pattern (Pai et al., 1997; Maki et al., 1997; Maki et al., 2003). When one foot is lifted during the single support phase, the base of support is roughly the space between the lifted foot and the other foot placed over the ground. This space increases when both feet contact the ground during the double support phase. Using a walking stick increases the base of support and stability while walking, and in doing so, regulates the centre of mass displacement (Maki et al., 2003). A wider base of support created by the third contact point with the ground can help the walking stick user to avoid instability by moving the centre of mass within the base of support margins for longer in the duration of a gait cycle. Additionally, the use of a walking stick creates a bilateral stability that can provide better control of the horizontal (FCH) and vertical (FCV) components of the hand reaction forces leading to better maintenance of equilibrium, as these hand forces are exerted against the centre of mass displacement. The moment of instability during the toe-off (FW×A) is generated by the body loading on the opposite supporting foot, unbalancing the centre of mass towards the lifted foot. As demonstrated in Figure 11, the applied weight on the walking stick can counterbalance the load of the unsupported limb (FV=FW−FCV) (Bateni & Maki, 2005).

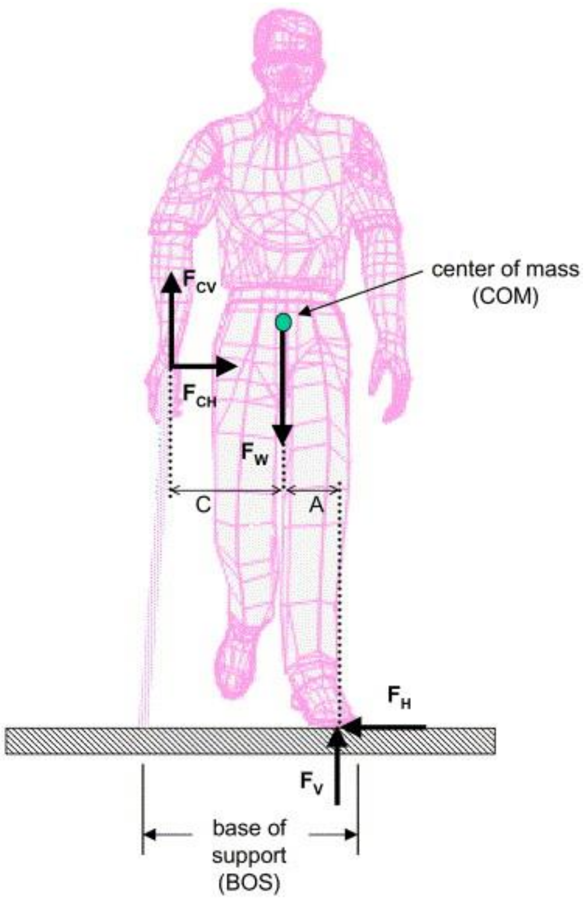

Fig 11. *Illustration of walking stick use increasing the base of support during toe-off. (Bateni & Maki, 2005).*

Walking sticks are used in a variety of ways, depending on the individuals' personalised needs or on the type of impairment they suffer from (Whittle, 1998). To date, despite the widespread usage of walking sticks, the number of studies investigating assisted locomotion in the elderly is relatively low (Jones et al., 2011). Previous studies have investigated the side on which the walking stick should be held (Murray et al., 1969; Klenerman et al., 1973; Ely et al., 1977; Edwards et al., 1986; Kuan et al., 1999; Stowe et al., 2010). Patients making ipsilateral use, that is, holding the walking stick on the same side as the affected limb, demonstrated an unstable gait pattern (Stowe et al., 2010). On the contrary, it was found that patients suffering from pain and weakness in one limb who held the walking stick on the opposite side, thus making contralateral usage, lightened the loading on the affected limb and counterbalanced their stability (Edwards et al., 1986; Stowe et al., 2010). Bennett et al., (1979) suggest that the use of a walking stick during gait can generate impulse forces in the antero-posterior direction. As demonstrated in Figure 12, when the walking stick is held vertically and a moment is exerted at the hand, propulsion is generated (A and B). If an opposite moment was exerted by the hand then a braking force would occur. As the moments exerted by the patients' hand are often minimal due to weakness, a more efficient way to create propulsion (C) and braking (D) is by modifying the angle at which the walking stick is held. This propulsion or braking force could improve the initiation or the deceleration of the impaired gait of elderly osteoarthritis patients. Until now, the use of walking sticks has been studied for hemiplegic stroke patients (Chen et al., 2001) and patients with spinal cord injuries (Melis et al., 1999) but less is known about osteoarthritis patients who most often have multi-articulate pain spread in different parts of the body.

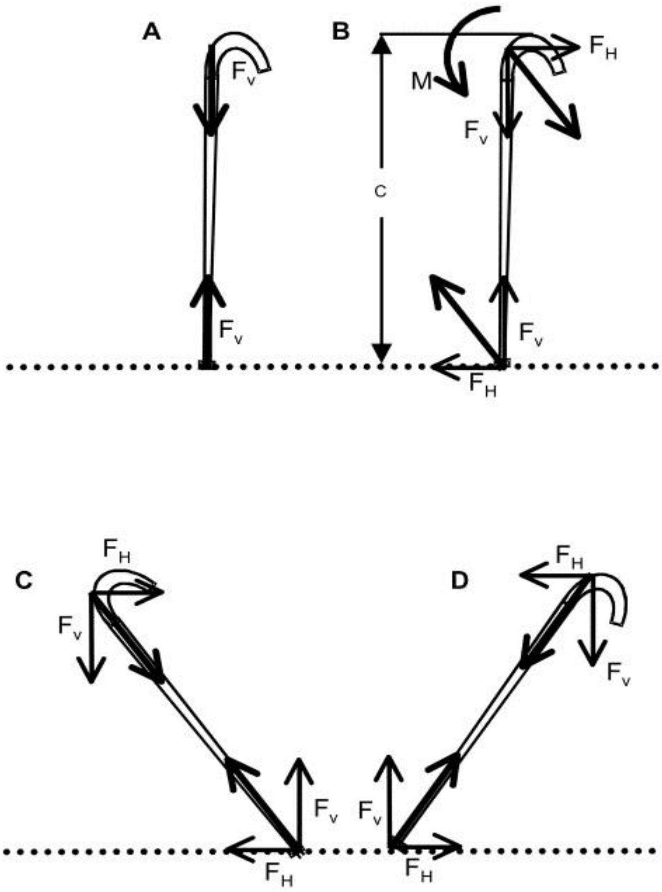

Figure 12. *Illustration of the applied hand forces on the walking stick creating horizontal ground reaction force and therefore propulsion (B, C) and braking (D) in assisted locomotion. When the walking stick is in vertical position a moment (M) need to be exerted in order to create propulsion (A,B) and vice versa an opposite moment is required to brake. The walking stick users regulate propulsion and braking by holding the stick in different angles (C, D) (Bateni & Maki, 2005).*

Using a walking stick may also have a positive psychological effect on patients' gait. Elderly walking stick users have reported feeling more confident and safer when walking and doing daily activities compared to other elderly patients who do not make use of this aid. The factors of confidence and safety can lead to a healthier lifestyle as the increased amount of physical activity implies independence and better general health (eg cardiovascular and respiratory function, blood circulation)

(Aminzadeh et al., 1993; Dean et al., 1993; Jaeger et al., 1989). Although the effects of ageing and osteoarthritis are separated from the walking stick use, they all affect human locomotion and influence the interaction of the elderly with the environment. This thesis will investigate the two critical points regulating the interaction of the elderly with the environment during the walking activity. These two critical points for are the contact with the ground i) by wearing shoes for those who experience difficulties to walk, and ii) by using the walking stick. The first chapter will focus on the effects of shoes during gait, and compare kinematic and kinetic variations in gait through the lifespan in natural locomotion, both in barefoot and shod conditions. In the second chapter I will investigate locomotor abilities in elderly osteoarthritis patients who use a walking stick to support gait. The aim of that chapter is twofold. First of all to use a questionnaire approach to analyse the reasons (including health conditions) that led participants to use a walking stick and secondly to obtain basic kinematic data of assisted locomotion in the same individuals in their natural environment, which can be linked to their responses in the questionnaire and thus to their experience of stick use. The third study will compare spatio-temporal parameters and kinematics of level walking by elderly walking stick users in indoor lab conditions and natural outdoor environments in order to understand locomotor variability and quantify the value of lab-based approaches to understanding gait in elderly patients, given the simplified locomotor environments present in clinical labs. Finally, the last experimental chapter will investigate the use of a walking stick combined with other aspects of the individual's locomotor environment such as sloped surfaces. The gait analysis in this chapter will report the differences in gait kinematics between level and sloped walking comparing gait between elderly osteoarthritis patients using walking sticks relative to other elderly individuals that don’t use a walking stick and a young, healthy group. My overall goal in these chapters is to look more deeply at the constraints that affect locomotion in elderly osteoarthritis patients and how these can influence their quality of life and risk of falling.

CHAPTER 2: COMPARISON OF BAREFOOT AND SHOD WALKING THROUGH LIFESPAN

2. Comparison of barefoot and shod walking through the lifespan

Study design (Maria Ntolopoulou, François Xavier Li); experimental design, ethics and data collection (Maria Ntolopoulou, Philip Hodgson [BSc Student], Helen Moss Student [BSc Student]); data analysis (Maria Ntolopoulou), data interpretation and chapter writing (Maria Ntolopoulou, Susannah Thorpe).

2.1 Introduction

It is well established that walking is a fundamental activity contributing to human health and wellbeing (Baker et al., 2008). The foot is a critical contact point regulating human interaction with the environment by providing sensory motor information to the body necessary for gait stability and movement (Kavounoudias et al., 2001). The human foot is a complex structure forming a longitudinal arch with 26 bones, 33 joints and 19 muscles to absorb the body weight and spread the exerted forces (Theodore, 2008; Mckeon et al., 2015). From an evolutionary perspective, barefoot walking is the natural means of locomotion so the feet of modern humans have had to adapt to walking in shoes, which provide constant support to the muscles forming the arch in the sole (Lieberman, 2010). The type of footwear, the texture and the shape of the shoe sole can affect gait spatiotemporal parameters and foot joint angle kinematics (Nurse et al., 2005; Menant et al., 2009). The fact that in developed countries not everyone wears the same type of shoes, that we start wearing shoes at a very young age and that we wear different types of shoes during our lifespan needs to be taken into consideration. When continuously reinforced by the shoe, the structure of the arch becomes weakened, similar to the effect of constant upward pressure of the keystone of a bridge arch weakening its structure, so that when footwear is removed, people look ‘flat footed’ (Lieberman 2012). This adaptation may be explained by alterations or dysfunctions in gait, such as a collapse of the arch of the foot (Van Boerum and Sangeorzan, 2003), reduction in proprioception, shortening of the gastrocnemius muscle fibres, and stiffening of the Achilles tendon (Csapo et al., 2010), all of which reduce the active range of motion possible at the ankle. From an anatomical point of view, it has been suggested that the chronic use of shoes from an early age may affect the shape and the biomechanical function of the forefoot (D’Aout et al., 2009). It has been suggested that the chronic use of high heels can cause alterations to the ankle muscle-tendon unit (Csapo et al., 2010). In fact, the thicker the shoe sole is, the more unstable the ankle would be due to the over solicitation of Peroneus Longus (Ramanathan et al., 2011) and the more the disruption that would be caused to natural motion (Stacoff et al., 1991)Moreover, several studies have suggested that barefoot walking is beneficial for health: according to Lieberman (2012), it can improve proprioception and muscle-tendon complex function, increase flexibility and mobility of foot mechanisms, improve balance, increase circulation, enhance overall foot health and could contribute to the prevention of injuries (Jenkins et al, 2010).

The literature suggests that gait spatiotemporal parameters differ between barefoot and shod walking, with shod walking being associated with increased step and stride length and walking speed, and decreased step rate, step width, stance phase duration, and double support time (Lythgo et al., 2009; Oeffinger et al., 2009; Wolf et al., 2008; Keenan et al., 2011). Furthermore, gait variations identified in joint angle kinematics show that walking with shoes can constrain gait, decreasing both the maximum plantar flexion angle and the maximum knee flexion (Oeffinger et al., 2009; Morio et al., 2009). The ankle position at contact of the foot with the ground also varied between barefoot and shod walking. It was suggested that in shod walking the touchdown has rather a heel strike pattern whilst in barefoot walking rather a flatter foot strike, and consequently a higher range of motion of the ankle joint in barefoot walking (Zhang et al., 2013). Additionally, in kinetics, the loading rate of the vertical ground reaction force was found to be less in shod walking compared to barefoot (Zhang et al., 2013). In fact, an amount of the force is absorbed by the cushion of the shoe sole. Therefore, the thicker the sole, the more the force that is absorbed as opposed to barefoot where there is no cushion under the toe (Zhang et al., 2013). The human foot at birth does not have the same structure as an adult foot. It is mainly a mass of cartilage which ossifies to become the 28 bones that are present during adolescence. This process usually occurs around the age of 13 years in girls and 15 years in boys (Evans, 2010). The walking development from the young age and through the course of lifespan can be particularly influenced by shoes with thick, stiff and rigid soles have no flexibility with result the restriction the forefoot movement (Staheli, 1991; Walther et al., 2008). Although the above gait variations between barefoot walking and different types of footwear have been recently reported in a systematic review (Franklin et al., 2015), to date there seems to be a lack of research on walking over the course of the human lifespan, with none as yet considering how the effect of shoes on gait might vary according to age. It has been shown that age-related changes affect gait parameters reducing speed, step length and step rate (Winter et al,.1990, Prince et al., 1971). Wolf et al. (2008) conclude that footwear significantly constrains the natural motion of children with regard to inward movement of the foot (inversion) and outward movement of the foot (eversion) of the subtalar joint motion during the stance phase of walking. Developing from birth and mastered by the age of three (Sutherland et al., 1988), bipedal walking allows independent movement but with ageing there is a decrease in physiological capacities and in physical activity levels, which could be attributed to a number of natural causes such as loss of balance or decrease of muscle mass (Reyes-Ortiz, Al Snih, & Markides, 2005). Previous research has shown that healthy elderly subjects had a significantly shorter step length and wider step width compared to young adults. Despite these findings which suggest that the range of motion in different joints is reduced with age when walking (Paroczai et al, 2006), age-related variations in growth and development through the lifespan between barefoot and shod walking are currently unknown.

Previous literature that compared the stance phase and the contact with the ground between barefoot and shod running (De Wit et al. 2000), also in populations that habitually ambulate barefoot or with minimal cushion in their shoes (Lieberman et al. 2010), have demonstrated that the type of footstrike is different between barefoot and shod running. It is therefore important to observe the foot position at the moment of touchdown in walking as well in order to provide a better understanding of the underlying mechanisms responsible for gait variations induced by shoes. The aim of this study was to explore kinematic and kinetic variations in gait between walking barefoot and wearing shoes and how these variations develop through different ages of the lifespan.

Based on previous studies which have suggested that shoes affect barefoot locomotion in children by reducing their foot mobility (Rao et al., 1992; Wolf et al., 2008) and others which have concluded that shoes can increase balance in the elderly (Horgan et al., 2009) - or that it might be recommended for the elderly to wear shoes even in indoor environments to prevent the risk of falling (Koepsell et al., 2004; Menz et al., 2006; Kelsey et al., 2010) - I hypothesised that the footwear effect would be different in young and old individuals.

Therefore, the natural locomotion pattern in the younger groups was expected to present a distinct difference between the two conditions: the foot mechanism should be more mobile in barefoot condition with higher plantar flexion and there should be more restricted ankle mobility and range of motion while wearing shoes (Wolf et al., 2008). It was expected that as we get older and become used to walking in shoes, the differences between barefoot and shod walking patterns would diminish, as the foot has adapted to footwear support, and if we take into account the fact the arch weakens, not surprisingly, older age groups would be more flat-footed and, most likely, feel more comfortable and have higher ankle mobility wearing shoes (Waddington et al., 2004). Moreover, ankle mobility should drop to lower levels because of the anticipated general decrease in physical capacities that comes with ageing. To understand whether the differences in gait patterns between barefoot and shoes are maintained at different speeds, we compared two self-selected speeds: comfortable and fast.

2.2 Method

2.2.1 Participants

Four experimental groups were formed (Table 1): children, young adults, middle aged adults and elderly, comprising a total of 37 participants. Of the total 19 male and 18 female subjects, all were healthy, physically active, aged between 10 and 80 years, free from physical impairment that would have prevented them from engaging in walking activity, and with no clinical history of disease or injury in the lower extremities or any other disability. The experimental procedure was approved by the ethics committee at the University of Birmingham.

	children	young adults	middle aged	elderly
n=	8 (3 female)	10 (5 female)	10 (5 female)	9 (5 female)
age	12±1	19±1	48±6	76±5
body weight	39.15kg±6.52	69.03kg±5.71	74.54kg±7.54	73.82kg±7.44
height	148.14cm±7.83	173.07cm±7.52	172.81cm±8.85	170.72cm±7.42
shoe weight	655gr±235.8	693gr±168.9	690gr±106.1	710gr±183.3

Table 1. *Summary table of participants' demographic characteristics (age, body weight, height and shoe weight mean/ standard deviation)*

2.2.2 Task and procedure

Participants were asked to walk along a 10m walkway and step on a force plate with their dominant foot. No instruction was given for the footwear, participants came with the shoes they wear in daily activities. The argument behind this was ecological validity, since we can observe the gait pattern occurred by their choice of footwear. Three trials of practice were given to regulate the pathway and step on the force plate. Each participant completed a total of 20 trials, with 10 trials at two self-selected speeds for each of the two different footwear conditions, i.e. barefoot and footwear. Five trials were conducted in each of the two self-selected speeds, namely, preferred (mean= 4.31 m/s, S.D=0.37) (participants instructed to walk at their preferred and comfortable speed), and fast (mean= 5.58 m/s, S.D =0.66) (participants instructed to walk as fast as possible). The order in which they completed the two footwear conditions and two speed conditions was determined by counterbalancing, so that an equal number of participants in each group started with each speed and condition, thus eliminating any potential effect that the order might have on the results. Before testing began, participants were weighed, and their shoe weight and dominant foot also noted. Kinematic data were collected with a 13 camera motion capture system (Vicon, Oxford, UK) at a sampling rate of 250 Hz while (Kistler) force plate data were sampled at 1000 Hz. Twelve markers were applied on the hips, knees and feet (heel, toe, ankle) (Figure 1).

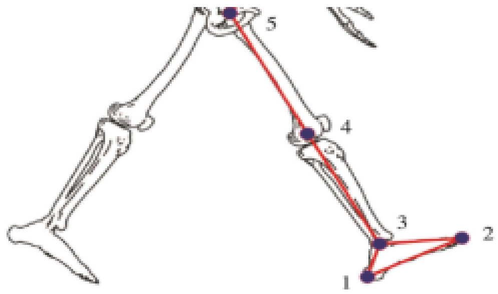

Figure 1. *Marker placement on the lower body 1) heel 2) toe 3) ankle 4) knee, 5) hip.*

Participants were asked to wear their own shoes (Figure 2), a decision which was considered to be more congruent with the ecological context, even though it could have an impact on variability.

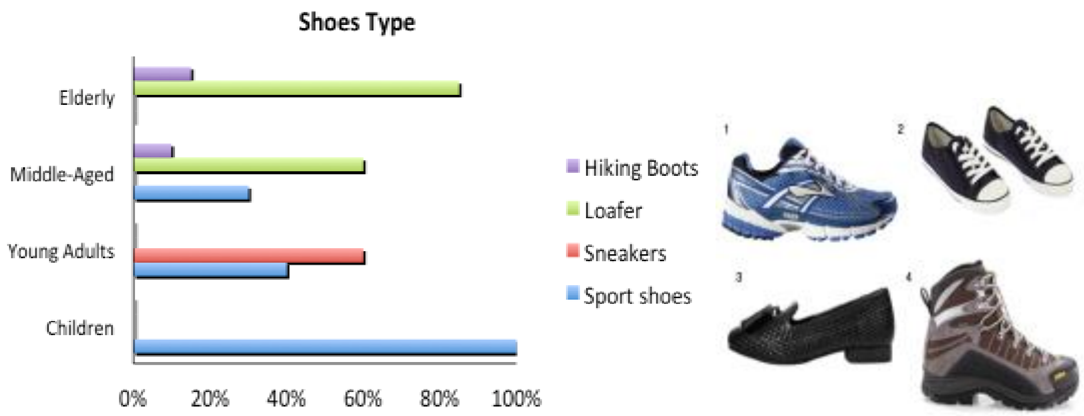

Figure 2. *Summary chart describing participants' footwear type used during testing 1) Sport shoes (flexible shoes over a thick platform sole used for sports activities), 2)*

sneakers (converse or similar with flat sole made of rubber), 3) loafers (rigid shoes with low heel), 4) hiking boots (ankle boots with thick sole).

2.2.3 Data analysis

The timing of foot strike and toe off, coincident with the first and last frames of ground reaction force data, was used to define the stance phase of each trial. Age category, gender, footwear condition and speed were defined as independent variables. The analysed gait spatiotemporal parameters of the gait cycle were the stance and swing phase duration, walking speed, cadence (steps/min), step length being the distance between the point of initial contact of one foot and the point of initial contact of the opposite foot. Kinematic data such as sagittal plane foot motion were measured by plantar-flexion/dorsiflexion (PF/DF) angle and calculated on Matlab. The stance phase, the time from heel strike to toe off, was normalised so that data could be presented as a percentage of the gait cycle. The force exerted could show the underlying cause of the abnormity or any alteration on the gait, which is the reason why we observed kinetics. Ground reaction force (GRF) is defined as the force exerted by the ground in reaction to a force applied to it and its results were normalized to body weight in order to compare measurements of subjects of different ages; the peak values of the vertical component of the force between the floor and the foot were calculated, as spatiotemporal, kinematics and kinetics were dependent on age, gender, footwear condition and speed.

2.2.4 Statistics

Multivariate analysis of variance (MANOVA) was used to test the effects of gender, age, speed and footwear condition on kinematic and kinetic parameters in SPSS v. 21, and significance threshold was set to p=0.05 after Greenhouse-Geisser correction in order to use a strict criterion of statistical significance. Greenhouse-Geisser corrects the violation of sphericity. In the case of this experiment, due to the small sample, the repeated measures violated the assumption of sphericity. This is why the Greenhouse-Geisser was chosen to multiply by one the sphericity estimates, make the degrees of freedom smaller and the f-ratio smaller as it has to be bigger to be deemed significant. This procedure was done automatically by the spss. All trials were included but because of the small sample size only the main effects and two-way interactions of the independent variables were reported. Where appropriate, Bonferroni post-hoc tests were used to assess differences between conditions.

2.3 Results

2.3.1 Spatiotemporal Gait Parameters

The MANOVA showed that absolute speed was influenced by the interaction between speed classification (spontaneous, fast) and age (F (3, 33) = 6.213, p <0.00, n^2=0.361). Participants' spontaneous speed was significantly slower than their fast speed (spontaneous mean: 1.2 m/s (S.D. = 0.28) vs. fast mean 1.55 m/s (S.D. = 0.66) m/s; p <0,001). The lowest speed was recorded in the elderly group and when participants were asked to walk spontaneously they moved faster in shod condition (1.16 m/s S.D. =0.12) rather than barefoot (1.04 m/s S.D. =0.11). However, when participants were asked to walk as fast as possible, no significant difference was detected between the two conditions.

Step width was influenced by speed and the interaction between speed and age (F (3, 33) = 38.616, p=0.000, n^2=0.778). Participants' step was wider at spontaneous walking speed (20,99 cm (S.D. = 3.75) vs. fast 22.3 cm (S.D. = 5.83)). Step width increased with age with the elderly group presenting the widest step in all conditions.

Step length was influenced by the main effects of speed and footwear and interactions between age and speed ($F (3, 33) = 41.438$, $p < 0.001$, $n^2=0.790$) and age and footwear ($F (3, 33) = 63.080$, $p < 0.001$, $n^2=0.852$). Step length was higher at fast speed (barefoot 446.4 cm (S.D. = 76.47) vs shod 463.56 cm (S.D. = 90.01) than at spontaneous (barefoot 423.41 cm (S.D. = 72.97) vs shod 443.01 cm (S.D. = 90.01).

Step rate was influenced by speed, and interactions between age and speed, age and footwear ($F (3, 33) = 254.364$, $p < 0.001$, $n^2=0.505$), and speed and footwear ($F (3, 33) = 147.513$, $p < 0.001$, $n^2=0.273$). Participants' spontaneous step rate was significantly slower than their fast (spontaneous: 122.55 st/min (S.D. = 14.2) vs. fast: 106.89 st/min (S.D. = 14.12)). The lowest step rate was found in the elderly group in spontaneous shod walking (87.16 st/min (S.D. = 3.54).

The stance phase was influenced by the main effect of age ($F (1, 33) = 12.088$ $p<0.01$, $n^2=0.486$) and the interaction between age and footwear ($F(3,33)= 8.92$ $p<0.01$, $n^2=0.594$). Speed was not found to influence this variable. The duration of the stance phase during the gait cycle increased with age and was higher in shod walking than in barefoot walking for all age groups.

2.3.2 Ankle Joint Angles

The maximum ankle angle was in dorsiflexion for all participants. MANOVA found that the maximum ankle dorsiflexion (DF) angle was influenced by the main effect of age ($F (1, 33) = 10.090$ $p<0.01$, $n^2=0.478$) and the interaction between age and footwear ($F(3,33)= 9.84$ $p<0.01$, $n^2=0.472$). Speed was not found to have any influence on (DF) angle. As showed in Figure 3a, in general, the maximum DF angle decreases with age. Children and young adults were found to exhibit higher maximum DF angles in barefoot walking (91.35° (S.D. = 4.28) and 86.05° (S.D. = 5.04)) rather than in shod walking (84.67° (S.D. = 4.43) and 80.42°(S.D. = 5.29)), but this pattern was reversed in middle-aged and elderly adults.

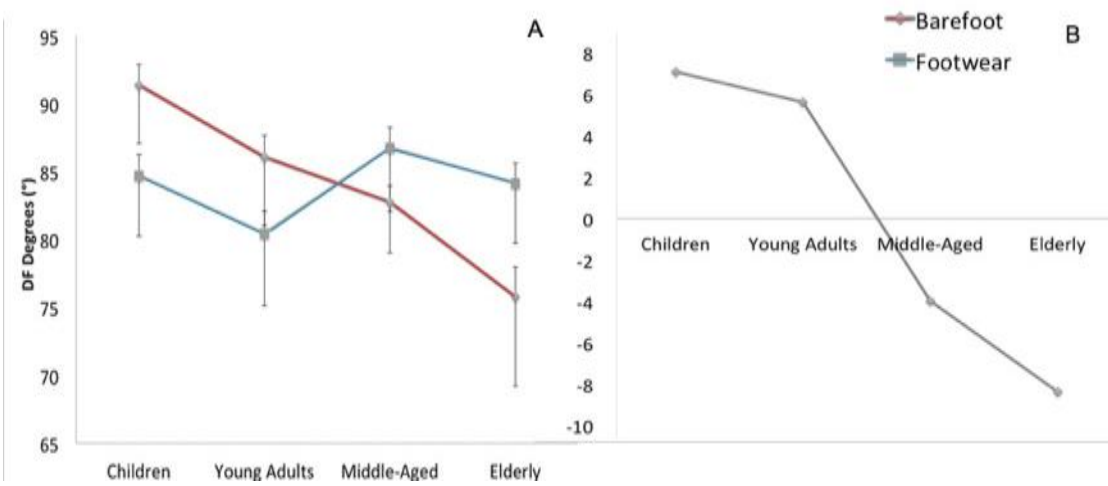

Figure 3. *A) Maximum dorsiflexion ankle angle and the range of difference between barefoot and shod walking for all four age groups. B) Difference between the two walking patterns given by the subtraction of footwear from the barefoot pattern in DF degrees through age.*

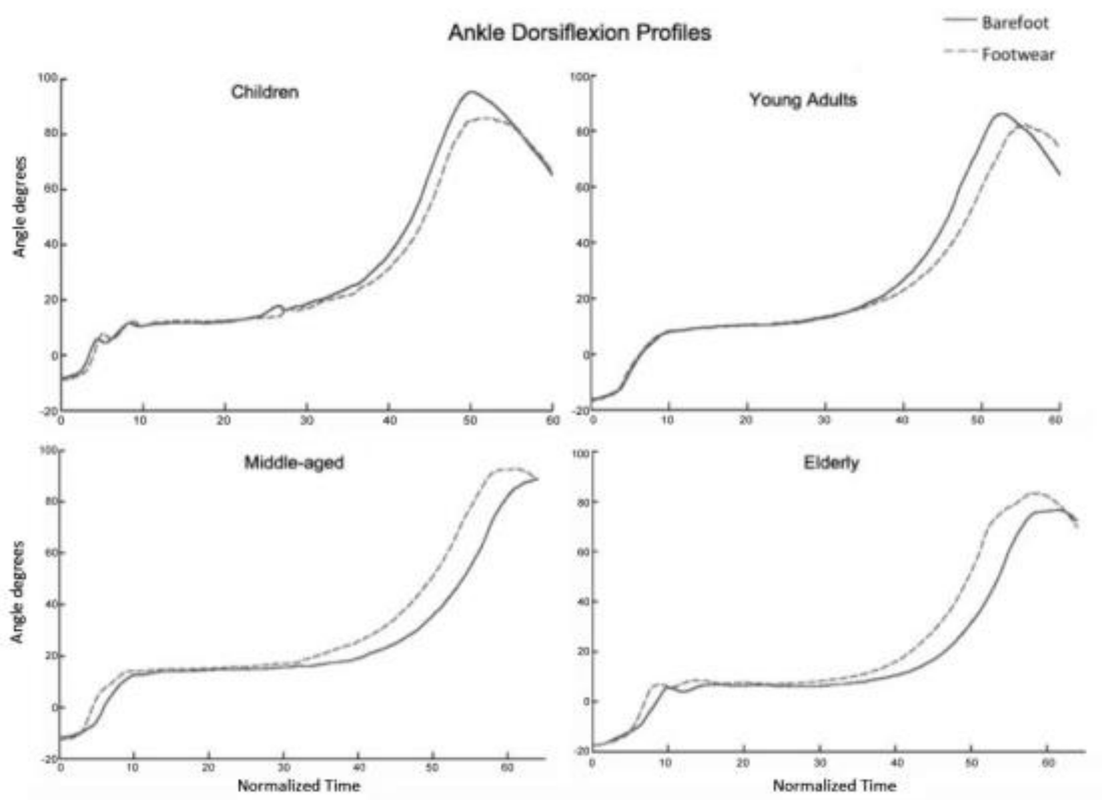

Figure 4. *Ankle dorsiflexion (angle degrees) profiles illustrating the patterns of barefoot and shod walking in normalized time (stance phase) for the four age groups.*

2.3.3 Kinetics

The MANOVA for maximum vertical ground reaction force (GRF) revealed that the initial peak impact of vertical GRF was influenced significantly by the main effect of age ($F (1, 33) = 43.884$ $p<0.01$, $n^2=0.571$) and the interaction between age and speed ($F (3, 33) = 5.825$, $p=0.003$, $n^2=0.346$). Figure 5a shows that the maximum initial peak of vertical GRF for barefoot walking decreased with age. In the initial peak, more force was applied for fast speed than spontaneous.

The push off peak was influenced by the main effect of age $F (1, 33) = 38.732$ $p<0.05$, $n^2=0.539$ and the interaction between age and footwear ($F (3, 33) = 58.289$, $p<0.05$, $n^2=0.346$). As shown in Figure 5b, the push-off peak decreased significantly with age for barefoot walking, while increasing slightly for shod walking only in the elderly group. The elderly applied the same amount of force (1.2 %BW (S.D. = 0.12)) for both barefoot and shod walking.

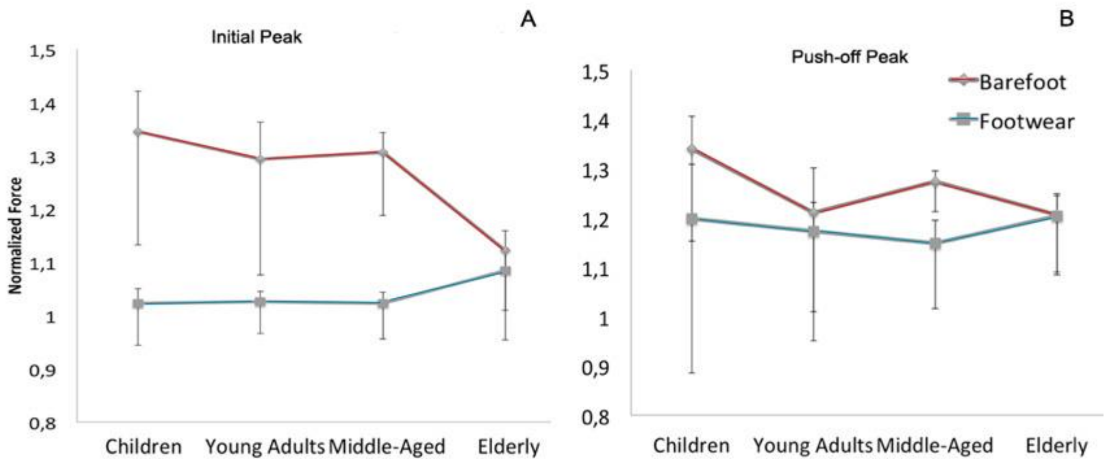

Figure 5. *A) Initial and B) push-off peak of the vertical ground reaction force normalized to body weight for barefoot and shod walking for the four age groups.*

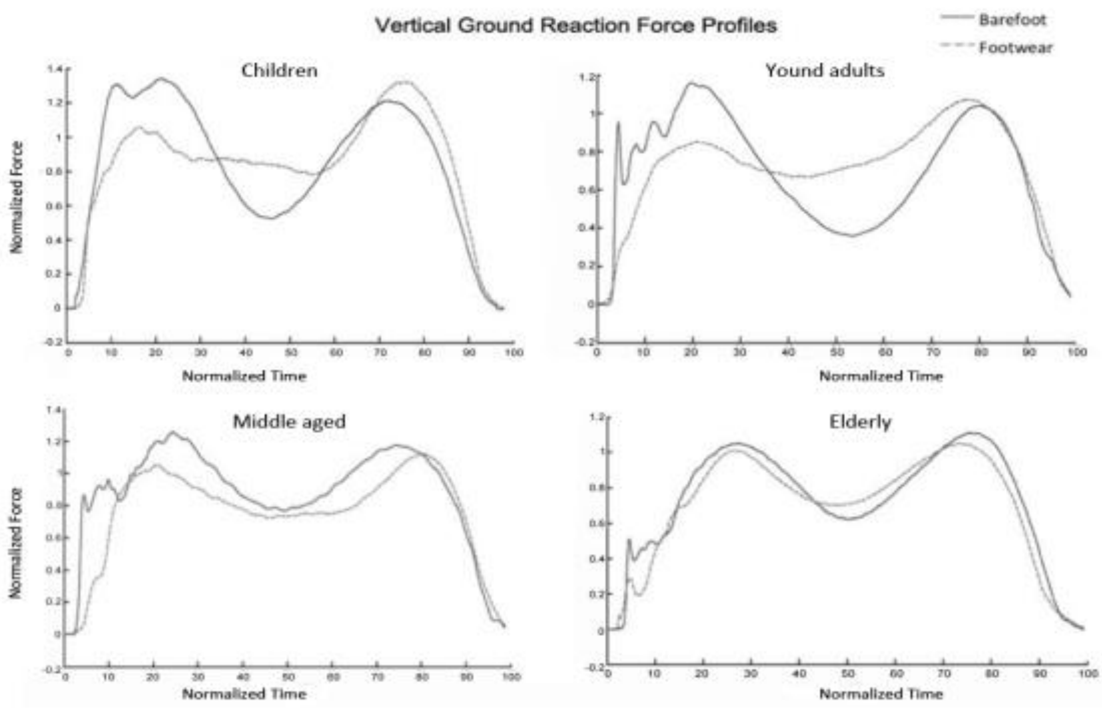

Figure 6. *Vertical ground reaction force (normalized force to body weight) profiles illustrating the patterns of barefoot and shod walking in normalized time for the four age groups.*

2.4 Discussion

The main objective of this study was to investigate kinematic and kinetic variations in gait between walking barefoot and wearing shoes and how these variations develop through different ages of the lifespan. The study was done with participants who are not used to walk barefoot in order to represent the average people of western societies who wear shoes since very young age. Based on previous studies by De Wit (2000) and Lieberman (2010) suggesting that barefoot runners contact the ground with a fore foot strike whilst shoes runners tend to heel strike, we hypothesized that the first difference between barefoot and shoes walking could be the type of foot strike. Our results showed that this was not the case in walking. At the moment of the touchdown, the initial peak of the vertical ground reaction forces was higher in barefoot walking whereas this peak seems to be absorbed by shoes. In contrast with other studies (De Wit et al. 2000; Lieberman et al. 2010), the spike of the impact peak in the barefoot pattern can potentially indicate that participants contacted the ground with a heel-strike rather than a foot strike. Therefore, it is possible that as a result of being accustomed to shoes walking, we become used to a forceful heel strike that is reproduced even when we walk barefoot. Striking the heels too forcefully may increase the mechanical stress in the joint. As a result, barefoot walking can become less comfortable or even painful so the cushion of shoes seems to become necessary with ageing and with the time of exposure to footwear. This speculation may be in accordance with Menz et al, (2006) who suggest that shoes should be recommended for the elderly even indoors.

Furthermore, we hypothesised that the magnitude of the footwear effect would be different between young and older individuals. Results revealed that the highest PDF angle was found in children walking barefoot, these results bearing out Wolf et al., (2008). Ankle angle analysis presented a restricted motion of ankle joint mobility while wearing shoes, as the highest plantar flexion was found in barefoot condition in children and young adults. In fact, in children we can observe two distinctly different patterns, with the barefoot natural locomotor pattern presenting the maximum ankle joint mobility which decreases while wearing shoes, and which might imply that movement is restricted by shoes. The reversal of the shod walking pattern (higher plantar flexion in the middle aged and elderly groups compared to younger individuals) could suggest that our movement is modified across the lifespan: in agreement with previous studies (Koepsell et al., 2004; Menz et al., 2006; Kelsey et al., 2010; (Waddington et al., 2004), it is not surprising that the middle-aged and the elderly presented a higher range of motion while wearing shoes and that a substantial difference between the barefoot and shoe pattern was reported for the elderly. This reversal, in congruence with Menant et al. (2009), could potentially corroborate the statement that wearing shoes and being supported by the cushion is a condition that we become accustomed to, as the foot is more mobile in this condition only for the middle-aged and the elderly. It was observed that the difference of the subtraction between footwear from the barefoot pattern was reduced in the elderly – an interesting finding which also suggests that with age we become accustomed to shoes.

Results also revealed a slower paced walking pattern in both barefoot and shod walking, as widely reported in the literature (Winter et al., 1990, Prince et al., 1971): slower spontaneous and fast speeds, an increase in step width while step length and cadence were reduced in the elderly, perhaps showing in this way a tendency to stabilise their gait pattern and find better balance while walking with or without footwear. A decrease in maximum plantar flexion in barefoot walking was also shown, an indication that in the elderly the overall range of movement available at the ankle joint is reduced. This decreasing level of plantar flexion with age in barefoot walking suggests that we become more 'flat footed' as we grow older. Another explanation could be, in congruence with Lieberman (2010), that the older we become, the more time we have had to adapt to wearing shoes and most likely, because the muscles forming the natural arch of the feet have weakened due to the constant reinforcement given by footwear.

As demonstrated in previous studies our kinematic results suggested that generally participants walked faster in shoes, taking slightly longer steps (Fanklin et al., 2015). Kinetics confirmed that footwear absorbs part of the vertical force applied to the ground. The lower peak found in shoes condition is, in actual fact, the result of the cushioning effect provided by shoes during walking. This result is in congruence with Zhang et al., (2013) who found a higher impact force in barefoot walking than in shoes. Although some people do not present an initial impact spike while walking (barefoot or in shoes), in our study it can be observed from ground reaction forces profiles that initial impact spike is lower or tends to disappear in the footwear pattern.

It is necessary to point out that this study has some limitations and that research regarding the influence of different shoes types and barefoot walking is scant. During the testing session of this study, subjects were asked to choose their own shoes and it was observed that children and young adults mostly chose flexible shoes whilst the middle aged and elderly wore stiffer and more rigid types. In order to reduce the variability arising from the different shoe types that participants wore, we could provide them with the same type of shoes during a familiarisation period before testing began. Furthermore, it was considered a challenge to expand the research to muscle activity and observe the muscular pattern in order to investigate the issue of mechanical stress. According to Murley et al. (2008), differences exist in EMG activation between various shoe conditions. Divert et al., (2005) compared barefoot and running shoes using EMG and reported significant differences in impact peak and impulse.

2.5 Conclusion

The main results of this study indicate that both kinematics and kinetics are influenced by the footwear and the age effect. In short, spatiotemporal gait parameters show that speed, step length and step rate decrease with age whilst step width increases. Kinematics gait analysis reveals that barefoot and footwear patterns differ and this difference, expressed in angle degrees, decreases with age. Children and young adults have higher PDF while walking barefoot, in contrast to middle-aged and elderly groups who display an inverse pattern with higher PDF in shod condition. Finally, kinetics reveal that the initial and push-off peaks fall with age, greater ground reaction force is applied while walking barefoot, and that the difference between barefoot and shod condition narrows with age.

Taking the above into consideration, it is our conviction that this study opens interesting perspectives regarding the influence of shoes on gait pattern. It would be interesting to investigate the impact of shoes on the mobility, balance and independence of the elderly population by analysing the type of shoes which would improve gait parameters, balance and stability during walking. Finally, it would be useful to look at the extent to which barefoot walking could be beneficial for the elderly and to investigate how factors such as proprioception, muscle strength, circulation or increasing the mobility of the foot could potentially improve balance during gait to prevent injuries or falls.

CHAPTER 3: ASSISTED LOCOMOTION IN THE ELDERLY: A CROSS-SECTIONAL STUDY TO UNDERSTAND THE CONTEXT AND MECHANICS OF WALKING STICK USE IN OLDER INDIVIDUALS LIVING IN A RESIDENTIAL CARE SETTING

3. Assisted locomotion in the elderly: A cross-sectional study to understand the context and mechanics of walking stick use in older individuals living in a residential care setting

Study design (Maria Ntolopoulou, Susannah Thorpe, François Xavier Li); experimental design, ethics and data collection (Maria Ntolopoulou); data analysis (Maria Ntolopoulou), data interpretation and chapter writing (Maria Ntolopoulou, Susannah Thorpe).

3.1 Introduction

Through the human lifespan gait patterns change and mobility problems increase with age (Bradley et al., 2011). Most of the changes in gait patterns that occur during adulthood are due to medical conditions such as osteoarthritis causing joint pain and deformities (Verghese et al., 2006). However, with ageing more attention is also required to perform the complex task of walking (Bridenbaugh et al., 2011). Increased attention is a cognitive age-related problem stemming from the general cognitive decline in ageing (Lezak, 1995). There is evidence to suggest that gait impairments in the elderly are linked with alterations in executive functions and attention (Yogev-Seligmann et al., 2008). For instance, previous MRI studies have found frontal and parietal activity during walking in elderly Alzheimer's patients (Malouin et al., 2003; Sheridan et al., 2007). In fact, neurological diseases or cognitive deficits can cause or aggravate gait impairments and increase risk of falling in older adults.

It has been reported that at least 30% of elderly (>65 years old) osteoarthritis patients encounter difficulty with more complex gait activities such as walking for more than ten minutes or with negotiating stairs, and almost 20% of them needed an assistive device (such as a walking stick) to facilitate gait (Centre for Disease Control and Prevention [CDC], 2009). Elderly people living in nursing homes and other residential care settings are reported to have more severe gait impairments than the general population (Verghese et al., 2006). Gait and balance disorders can also lead to falls; 90% of which occur in individuals over 65 years old. Falls can cause significant loss of independence and decrease quality of life (Rubenstein et al., 2002; Rubenstein et al., 2006). There is evidence to suggest that elderly people living in health care settings also have more falls than the elderly living in their own home environment (e.g., Rubenstein et al., 1988).

Improving gait can clearly increase quality of life (Schmid et al., 2007). It is well established that walking is a crucial activity that contributes to the health and wellbeing of older people (Baker et al., 2008). Walking sticks are widely used by elderly adults and have been shown to have a supportive role on gait (Jones et al., 2012). Clinical and biomechanics studies confirm that walking sticks can improve balance and mobility because the base of support for the individual becomes wider, the load of weight bearing is shared and therefore balance is improved (Bateni et al., 2005). Walking sticks can also help walking stick users focus on the task of walking by freeing them from concentrating on balance (Allet et al., 2009). It has been further shown that these devices can increase joint range of motion, muscle strength, joint stability, coordination, and endurance during gait whilst reducing pain, all of which contribute to enhanced independent functioning in patients with rheumatoid arthritis (RA) and osteoarthritis (OA) (Rogers et al., 1992). Using a walking stick for instance can reduce the load on the knee by 10% and by reducing knee joint stress, pain, swelling and stiffness can also be reduced (Simic et al., 2011). Walking sticks also seem to be beneficial for maintaining an upright posture during gait, which can increase autonomy, confidence and safety. Liu et al., (2011) analysed the gait of elderly walking stick users and found a significant association between the individual's posture during walking and their rate of falls, in that those with a more forwarded-leaning posture were found to have a higher rate of falls than those walking with a more upright posture. Finally, it has been also suggested that patients can increase their walking distance by 10% when using a walking stick (Comer et al., 2010).

Conversely however, walking stick use can also have a negative effect on the kinematics of gait, with some studies showing a reduction in speed, decrease in step length and increase in relative duration of stance phase (Liu et al 2009). It has been estimated that 12% of fall injuries in elderly people reported in the US emergency departments were associated with walking sticks (Stevens et al 2009). According to the systematic review of Bateni (2005) on assistive devices in spite of the importance of this topic relatively few studies have examined the biomechanical principles related to walking sticks use and the link between walking sticks and fall history (Bateni and Maki 2005, Liu et al 2009). This is largely because most studies have focused on patients after they have suffered a stroke (Corner et al., 2010; Chaminat et al., 2012) with the aim of evaluating the rehabilitation process (Tyson et al., 2009). In particular, little work appears to have been carried out on the long-term effects of walking stick use in daily activities by aging adults with osteoarthritis. One study that did examine a large sample of arthritis patients over 70 years old demonstrated that patients who experienced musculoskeletal pain in two or more locations, or reported pain during their everyday activities, were at a significantly higher risk of falling than those with no pain or with low levels of pain (Suzanne et al., 2009).

A further confounding factor with understanding the mechanics and efficacy of walking stick use is that to date there is not a well-defined procedure for the prescription of walking sticks, and usually the decision to obtain one and the structure of the stick is a patient's personal choice or based on a potentially subjective evaluation of a specialist shop keeper (Laufer, 2004). Liu et al., (2011) reported problems from three areas related to walking stick use that could increase the risk of falling:1) the lack of medical consultation and prescription can lead a large proportion of walking stick users (30% to 50%) to abandon their device soon after receiving it (Bateni et al., 2005); 2) the difficulties encountered during walking stick use and inadequate training in walking with the stick can aggravate the problem (Andersen et al 2007, Bateni et al 2005, Stevens et al 2009) and 3) the lack of fitting (the appropriate height adjustment adapted to the persons anthropometric characteristics) and cane maintenance (type and condition of handle and the stub) can lead to a reduction of its usage and therefore the amount and quality of walking may be lower than the person can physically achieve (Scherer et al., 1998).

Previous studies have suggested that walking speed can be an indicator to evaluate locomotor abilities (Salbach et al., 2001; Studenski et al., 2003). According to Schmid et al., (2007) the ability to walk fast is related to the perception of a higher quality of life. In the same study it was suggested that a gait speed of 0.8 m/s can be satisfactory for the elderly whilst the use of a walking stick can improve this performance by approximately 10% and these speeds can be considered as comfortable and fast.

However, achieving high walking speeds doesn’t necessary mean that the gait performance is high or that the walking stick is used in an effective way (Polese et al 2012). Testing the gait speed as the only parameter is not enough to characterize the locomotor pattern and understand of the nature of gait deficiencies.

Gait variability (i.e. variance of the gait parameter around the mean), is another parameter that can characterize instability during motion. Several studies have demonstrated that the high variability in gait spatiotemporal parameters can be related to a greater risk of falling (Hausdorff et al, 2001, 2007; Brach et al, 2005; Maki, 1997), which suggests that it may be able to serve as a tool to assess a participant’s potential risk of falling in the near future (Toebes et al.,2012). Dingwell and Cavanagh (2001) suggested that osteoarthritis patients and people with other gait disorders usually present greater variability in their walking pattern. Chronic pain can be a factor increasing variability in gait, affecting mobility and the daily functioning of the elderly (De Kruijf et al., 2015). According to a recent systematic review (Hamacher et al., 2015) gait variability measured through biomechanical parameters of foot kinematics can be a reliable way to quantify the stability of gait.

To date most research about the gait of walking stick users has been carried out with clinical gait analysis (Bateni et al., 2005; Polese et al., 2012) or with questionnaires (Wieczorowska-Tobis et al., 2013; Wallis et al., 2015). While these provide valuable information, it is important to link both measures of how stick users move around their natural habitats with how the walking stick is used relative to the individual’s recent (i.e. within a year) fall history. This chapter therefore contains 2 parts: a questionnaire (study 1) that aimed to collect information about why the elderly use their walking stick, how long they have been using it and why it was recommended they use it. Then a gait analysis (study 2) where participants were filmed with a camera while walking in their nursing home area to obtain baseline kinematic data on their natural assisted-gait. The overall aim was to associate walking performance to the data collected from the questionnaire, in an attempt to link gait parameters with how walking sticks are used, health condition and falls history. To date there are few published data on the locomotion of elderly participants in an ecological context and there is a need for further research in the understanding of the biomechanical requirements of walking with sticks and the design of instructions for practitioners, carers and users. The general purpose of this research was to provide new and global knowledge on walking stick use in the elderly.

3.2 Study design

Walking stick users over 65 years old, all residing in nursing homes in Birmingham completed a general questionnaire about their walking stick use, their health condition, their prescription or choice of walking stick and how, why, and by whom the walking stick was provided, fitted and followed up. Based on Rubenstein (1988) I decided to test elderly people living in health care settings as they have more falls than the elderly living in their own home environment. The questionnaire (appendix 1) also aimed to establish their level of satisfaction with their walking stick and parameters that they would like to improve. It was hypothesized that pain, disability (such as rheumatoid arthritis and osteoarthritis), fear of falling, need of general support and better balance would be the leading factors in obtaining a walking stick. It was expected that participants who are satisfied with their walking stick would use it more. Finally, we also expected a general lack of medical involvement in the obtaining and fitting procedure. However, we also hypothesized that, where a medical prescription or medical consultation had occurred, participant's would show an increased level of satisfaction with their walking stick and consequently longer use and better gait performance. Therefore, the key research questions for this questionnaire were:

1. What was the primary reason or health condition that led participants to use a walking stick and what was the extent of medical involvement in this choice?
2. What were the functional requirements of the walking stick during everyday activities and how satisfied were participants with their device?
3. What were the selection criteria and the fitting procedure of the walking sticks?

3.3 Method

3.3.1 Participants

Forty-seven elderly participants from different nursing homes in Birmingham using a walking stick in everyday life were recruited for this study: 26 were female (78.8 years ± 9.41; mass= 65.94 Kg ±.7.21, height= 159.85 cm ± 4.93) and 21 were male (77.28 years ± 8.7; mass = 76.17 Kg ±=9.36, height= 174.2 cm ± 7.65). Based on a previous study by Thaler-Kall et al, (2015) that found significant differences within gait parameters when the cohort was categorized by age (cut point 75 years) our sample was separated in three age-related sub-groups to reflect the different stages of ageing (65-74 years (N = 19; 69.36 years ± 3.54; mass = 68.94 Kg ±=7.34, height= 170.87 cm ± 7.93), 75-84 years (N = 19; 80.05 years ± 2.9; mass = 66.64 Kg ±=8.09, height= 167.32 cm ± 6.22) and over 85 years (N = 9; 92.77 years ± 3.01; body weight= 63.18 Kg ± 8.53, height= 164.12 cm ± 4.67).

3.3.2 Task and procedure

The questionnaire was created on Adobe Forms Central and data collection was performed on a tablet (Apple Ipad 2). The questions and options were read to the participant so that they simply had to answer the questions verbally. Data inputting was carried out in real time by MN and participants were able to see what was written at all times. The experiment was conducted in different elderly nursing homes in Birmingham (UK). The participant's stick height and the angle of the handle were also measured with a tape measure and a manual goniometer. Anthropometric data (weight, height) were collected and used for calibration in the vicon system and the clinical health history of each participant was recorded after obtaining their oral and written consent.

3.4 Results

The questionnaire was composed of three sections. The first section collected information about the health conditions that led participants to obtain their walking stick, the type of disability, medical recommendations and rehabilitation programs that the participant had been involved. Furthermore, this section investigated the participant's clinical history, the number of falls they had suffered in the last 12 months and their medical record of previous surgery. The questionnaire revealed that osteoarthritis (OA) was the main cause of disability in the elderly sample (83%). The affected joints were the knee (47%), hip (22%), ankle (12%) and the lower back (2%) whilst 11% experienced pain in the lower back and 6% experienced pain in both legs. Improving balance and compensating for osteoarthritis pain were the reasons leading the osteoarthritis patients (83%) to obtain a walking stick. The fear of falling (7%) or having had knee (7%) and hip replacement (3%) surgery were the other reasons leading participants to obtain a walking stick (Figure 1). For 59% of respondents it was a personal choice to use a walking stick whereas 24% were recommended by a GP and only 13% by an orthopaedic specialist or a physiotherapist. Seventy five percent of women and 63% of men of the tested sample had experienced at least one previous fall in the last 12 months. However only 27% of them followed physiotherapy and rehabilitation programs.

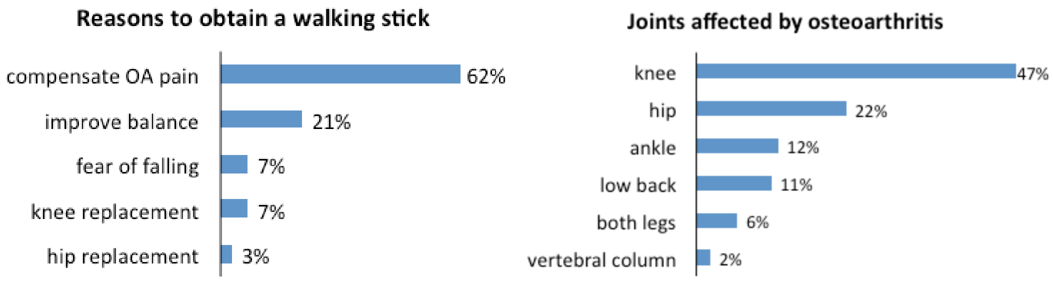

Figure 1. *Chart summarising the reasons that led participants to obtain a walking stick (left) and the joints affected by osteoarthritis (right).*

The second section focused on use of their walking stick, examining biomechanical requirements (i.e in which hand they hold their walking stick, walking distance) and participant's evaluation of their need for a stick, and their comfort and pain related to their device. Two items were used to assess the level of necessity for indoor and outdoor use of the walking stick ("How necessary is it for you to use your walking stick?) with answers scale ranging from (1) not at all necessary to (5) very necessary) and the level of comfort from it with answers scale ranging from (1) not comfortable at all to (5) very comfortable), respectively. The questionnaire found that a large proportion of the participants (93%) used only their walking stick to support gait whereas 7% used also a Zimmer frame. Most of the participants had started using their walking stick 5 years ago (43%) or longer (19%), whilst 22% obtained it 2 years ago and 16% within the last year after a fall accident. The majority held the stick in the opposite hand to the affected limb (69%) whereas only 31% held it in their dominant hand and 12% in both hands. Sixty-seven percent of participants used their walking stick every day, although 85% were able to walk without it and 11% declared themselves able to walk without it but for not more than 100m. All participants used their walking stick for outdoor walking and the everyday activities such as shopping, travelling or in the garden. However only 28% used it regularly in home (Figure 2). Using the walking stick was estimated to be more necessary in outdoor activities such as shopping, walking in the city centre and under difficult weather conditions (a score 4.6 out of 5) than in the home (2.8 out of 5) with scale ranging from (1) not at all necessary to (5) very necessary. On a daily basis 60% of participants used their walking stick at least 4-5 hours, 24% used it up to 3 hours and 16% up to 2 hours per day (Figure 2).

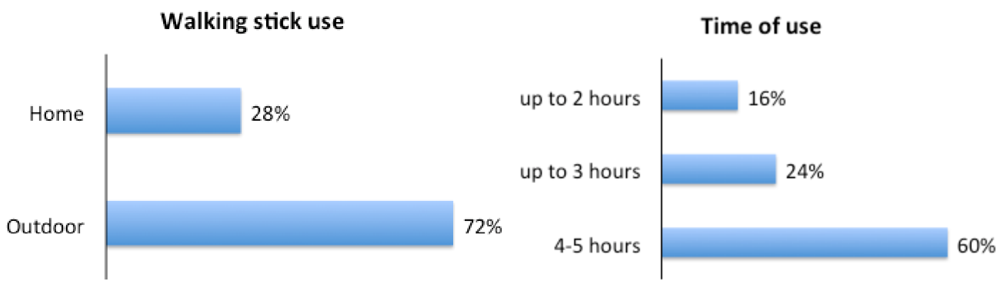

Figure 2. *Location and time of participants' regular daily walking stick use*

Using the walking stick was described as uncomfortable for 45% of the participants giving a score of 2.3 out of 5 according to a scale ranging from (1) not comfortable at all to (5) very comfortable. Furthermore 76% found that using the walking stick caused additional pain after a while, not only in the already affected joints but also in the participant's fingers (21%), shoulders (11%), hands (7%), neck (6%) and their lower back (14%) (Figure 3). The amount of time before participants experienced pain when using their stick varied depending on the type and the intensity of the activity. The simple task of walking was found to cause extra pain after 1 hour for 20% of participants and just 30 min for 9%. However, a more immediate effect of pain was found in activities with higher intensity; pain occurred instantly or in the first 30 minutes when carrying extra weight (54%), walking on slopes (36%) and negotiating stairs (10%).

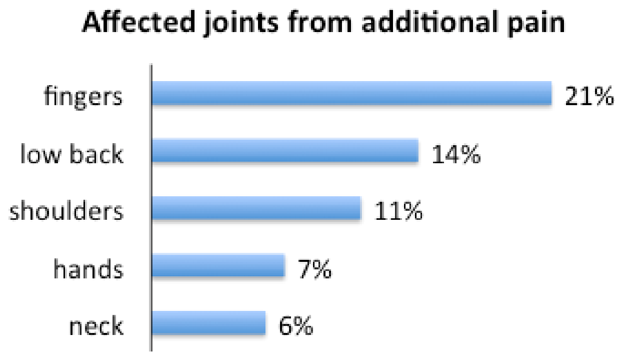

Figure 3. *List of the joints affected by pain caused by walking stick use.*

Finally, the third section examined the fitting procedure, the device's selection criteria and collected suggestions for improvements of the walking stick. In the fitting procedure, only 10% of the total provided sticks were fitted by an orthopaedic specialist or a physiotherapist, whist 48% were fitted by the shop or the manufacturer and 31% by the participants themselves (Figure 4). Among the selection criteria (Figure 4) for their walking stick were the adjustable height (36%), the design (14%) and shop/manufacturers suggestion (36%).

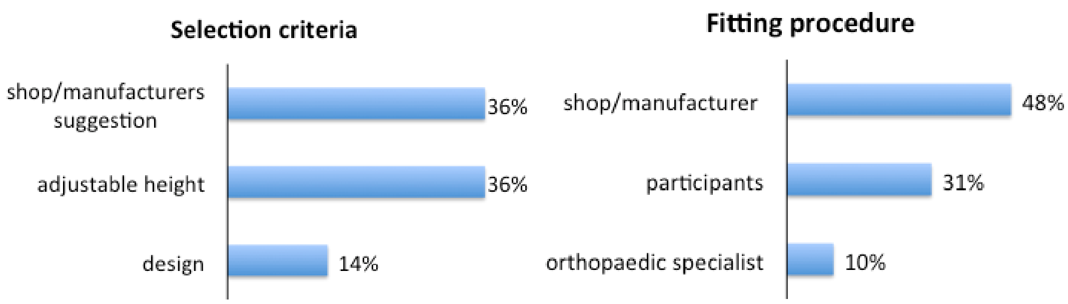

Figure 4. *List of participants' selection criteria for their walking stick and their fitting procedure.*

The walking sticks used in our sample were mainly wooden with fixed height (Figure 5D) (39%), height-adjustable made from aluminium (Figure 5B) (36%), carbon fibre with fixed height (16%) and carbon fibre with adjustable height- (9%) (Figure 5C). The handles were moulded plastic (36%), crook (39%), or with standard angled handled rigid shaft (19%). Only 6% had an anatomical grip (Figure 5A). The parameters that the walking stick users would like to improve were mainly a softer handle (68%), wider range of degrees of adjustment for the height (22%) on the stick and a more solid ferrule.

Figure 5. *Illustration of different types of walking sticks used in our experiment: (A) height-adjustable made from aluminium anatomical grip, (B) height-adjustable made from aluminium with standard rigid handle, (C) carbon fibre with height-adjustable and crook handle, (D) wooden with crook handle.*

3.5 Discussion

The first aim of this questionnaire was to identify the reasons and the health conditions that led participants to use a walking stick. Our findings show that walking sticks are selected not only by people suffering from a severe disability, who have had a previous surgery (hip, knee replacement) or a stroke as widely studied in literature. We found there is a large number of elderly individuals who suffer from moderate gait impairments that have decided to use a walking stick to compensate for osteoarthritis pain; for better balance and support during walking or as a result of a fall experience and/or fear of falling. In fact, 75% of women and 63% of men of the tested sample had experienced at least one previous fall in the last 12 months. This study also highlights the lack of medical involvement in the selection, use and fitting of walking sticks. We found that among this population, despite the high percentage of people who experienced pain while using their walking stick or had at least one fall in the last twelve months, only a few consulted a specialist or followed rehabilitation programs. These results are in agreement with Patcharawan et al, (2014), who conducted a questionnaire study of the reasons for using a variety of walking aids. Their study suggested that a proportion of the elderly tested sample used a walking device, particularly when walking long distances, due to a fear of falling, musculoskeletal pain, and impaired walking ability but only a very small percentage obtained it with a medical prescription. This is of concern since falls in elderly are a public health issue as they can cause not only mortality but also non-fatal injuries that increase the number of hospital admissions (National Center for Injury Prevention and Control, 2015). This has a major impact for the health system by significantly increasing the cost of health care for the elderly (Li et al., 2003). Earlier studies have suggested that the causes underlying falls are multifactorial and result from the interaction between internal (physiological) and external (environmental) factors (Newton, 2003). My results suggest that the lack of medical involvement in the selection and training in stick use may confound the associated health cost.

The second aim of this study was to analyse the functional requirements of walking stick use during everyday activities. We found that the walking sticks were used by all participants outdoors but only a few of them used it regularly in their home. According to the National Centre for Injury Prevention and Control the home can be also a hazardous environment having broken or uneven steps and objects that can be tripped over increasing the risk of falling. This risk is higher for those who have reduced ability of walking or experience musculoskeletal pain. Interestingly, we found that a large proportion of our sample declared that their walking stick caused them more pain after long use and not only in the already injured limb but also in other parts especially hands, arms and shoulders. This question needs further research to examine the processes by which holding a walking stick can expand pain to other locations. For instance, most of the participants were holding the stick in the opposite hand to the affected limb but there were also people holding it in the dominant hand or both hands. To date there is little scientific evidence to suggest whether there is a correct side to hold the stick relative to an affected limb, particularly when the individual may have osteoarthritis in their hands as well. Fang et al (2015) did examine holding the stick in the contralateral versus the ipsilateral hand and suggested that holding the walking stick either in the dominant hand or in the hand opposite the affected limb were both effective in reducing the load from the weaker side of the body.

The third and last purpose of the questionnaire was to examine the selection criteria and the fitting procedure of the walking sticks used in our study. We found that most of the provided walking sticks were fitted by the shop, the manufacturer or participants themselves. These findings are in agreement with the results of Liu et al., (2011) suggesting that only a small percentage of the elderly obtained a walking stick after medical prescription or recommendation, fitted correctly by a specialist. They are also in congruence with Patcharawan et al, (2014) who found that most of their participants fitted their walking stick themselves and only a very small proportion did so according to a medical prescription. Our findings highlight the lack of medical involvement in the decision to obtain and fit a walking stick, but a key problem is that they don't provide specific evidence on the optimal selection. This could be achieved with the analysis of the length of the walking stick in relation to the person's height, and testing whether there is difference between prescribed and non-prescribed stick use in gait kinematics. It would be interesting to carry on this research in the future to define what is the optimal selection process of a walking stick according to patient's anthropometric characteristics and individual needs.

Study 2

The aim of study 2 was to investigate how different factors identified in the questionnaire, including health, age and fall history are related to gait and to walking stick use. It is well known that functional capabilities of the human body declines with age, especially after the age of 75 (Branch et al., 2004). Patcharawan et al., (2014) recently suggested that the need to support gait using a walking device increased six times between people aged 60–74 years and those aged over 75 years old. Therefore, it was hypothesized that elderly in advanced age or those who had experienced a previous fall would demonstrate a reduced ability to walk naturally. Pain while using the walking stick was also expected to decrease gait performance. However, it was hypothesized that people who had been prescribed a specific walking stick by a GP or orthopaedic specialist would be able to use it to compensate more effectively for their gait disorders and walk better. Therefore, the research questions for the gait analysis were: 1. How do age, pain and falls affect the gait of elderly walking stick users in their natural environments? 2. Does medical involvement in the prescription and fitting of the walking stick improve gait performance?

3.6.1 Participants

The sample was composed of the participants from study 1(see Table 1) (N = 47; 26 women and 21 men) additionally categorized by falls history (previous fall in the last 12 months or not), pain (additional pain from using their walking stick or not) and medical recommendation (with or without prescription from GP/Orthopaedic specialist).

3.6.2 Task and procedure

Kinematic data for the gait analysis were collected by filming participants in a calibrated setting with two (non-synchronised) Sony Video Cameras mounted on tripods and recording at a frame rate of 25 Hz. Participants walked freely along a 10m stretch of their natural environment outside their nursing home. The 10m stretch was selected for its flat surface and absence of obstacles. The start and end points of their walkway were clearly marked. After practice trials to familiarize the participants with the procedure, each one completed ten walking trials, which were filmed from the lateral perspective (from the side the stick was held) and from a frontal perspective. Five strides were selected from the middle of the walking sequence for analysis. Gait kinematics were analysed with Tracker Video Analysis and Modelling Tool (version 7.3) using high resolution monitors (1920 x 1080 pixels) to display the maximum image definition. The reference points (head, sternum, shoulder, hand, heel, toe) were identified and their position was tracked automatically by the software instead of applying Vicon markers on the participants, as in Chapter 2. Protractors and tape measures integrated in the software provided distance and angle measurements. Any potential distortion was automatically corrected from the software with a radial distortion filter.

3.6.3 Data analysis

Age, gender, fall history, medical recommendation and pain were defined as fixed effects (Independent variables). Spatiotemporal parameters of the gait cycle were evaluated as dependent variables. Stance phase duration (%), walking speed (m/s), step rate (steps/min), step length (cm), width (cm) and the stick contact time with the ground were calculated. The trunk flexion and the elbow flexion - extension angles were calculated to assess whether participants used a forward leaning position during walking and were included in the gait analysis. Stance phase, the time from heel strike to toe off for a single limb, was normalised to 100% so data could be compared within and between subjects. Changes in gait variables within and between groups were tested with a multivariate analysis of variance (MANOVA) in SPSS v. 22. Between subjects factor are listed in table 1. The significance threshold was set to $p=0.05$ after Greenhouse-Geisser correction in order to use a strict criterion of statistical significance. Where appropriate, Bonferroni post-hoc tests were used to assess differences between conditions.

Between-Subjects Factors			**N**
Age	1	65-75	19
	2	75-85	19
	3	>85	9
Gender	1	Female	26
	2	Male	21
Medical Recommendation	1	With Recommendation	17
	2	Without Recommendation	30
Pain	1	With Pain	16
	2	Without Pain	30
Fall	1	With Fall	32
	2	Without Fall	15

Table 1. *Summary of fixed effects and size of tested sub-groups*

3.7 Results

Age

The MANOVA found that speed ($F (2, 46) = 22.327$, $p = 0.000$, $n^2=0.71$), step rate ($F (2, 46) = 13.091$, $p = 0.000$, $n^2=0.373$), stance phase ($F (2, 46) = 8.522$, $p = 0.001$, $n^2=0.279$) and step width ($F (2, 46) = 11.875$, $p = 0.000$, $n^2=0.351$) were all affected by the main effect of age (Figure 1). Pairwise comparisons showed that the >85 years group walked significantly slower (0.52 m/s ±0.16) than the 75-85 (0.86 m/s ±0.38) who also walked significantly slower than the 65-75 (1.38 m/s ±0.34). For step rate (>85 (75.82 step/min ±29.41); 75-85 (81.71 step/min ±25.93); 65-75 (96.59 step/min ±23.94)) stance phase (>85 (69.85% ±1.95); 75-85 (68.03% ±2.6); 65-75 (66.04% ±2.28)) and step width (>85 (37.27 cm ±5.01); 75-85 (33.2 cm ±5.19); 65-75 (28.97

cm ±2.81)) pairwise comparisons showed that the >85 years group was significantly different from the other two age groups, who were broadly similar to each other.

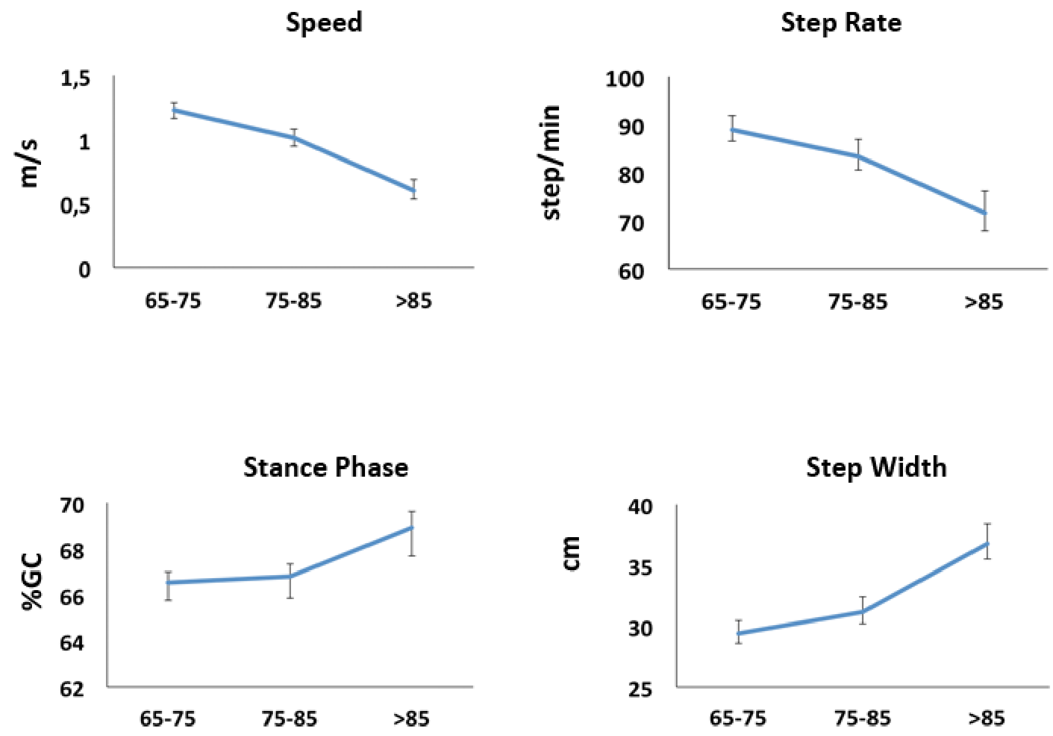

Figure 1. *The relationship between age group and kinematics speed (m/s), step rate(step/min), stance phase (percentage of gait cycle) and step width (cm) (95% confidence intervals in error bars).*

Gender

The MANOVA found that the maximum elbow flexion angle of the stick-using arm (F (1, 46) = 5.575, p =0.023, n^2=0.332) was the only kinematic variable affected by the main effect of gender. Women's elbow's angle was found to be maintained in a more flexed position than men's (78.72° ±22.91 vs 110.97 ° ±28.28) (Figure 2).

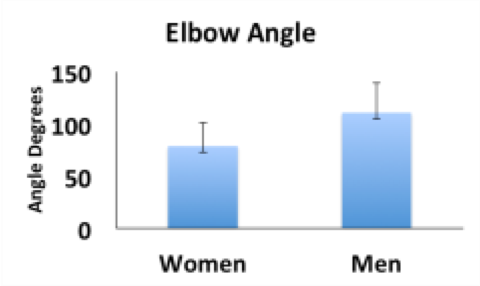

Figure 2. *The relationship between gender and maximum elbow flexion angle degrees (95% confidence intervals in error bars).*

Medical recommendation

The MANOVA found that speed (F (1, 46) = 22.327, p =0.002, n^2=0.114), step width (F (1, 46) = 18.733, p =0.000, n^2=0.294) and cane contact time with the ground (F (1, 46) = 13.771, p =0.001, n^2=0.234) were all affected by the main effect of whether the individual had obtained his/her walking stick after medical recommendation. Pairwise comparisons showed that those who obtained their walking stick after medical recommendation had a significantly (p<0,005) higher walking speed (1.21 m/s ±0.47 vs 0.88 m/s ±0.37), higher step width (34.4 cm ±5.01vs 28.51 cm ±9.6) and lower cane contact time (0.9 s ±0.41 vs 1.32 s ±0.33) than those that had obtained their device without any recommendation (Figure 3).

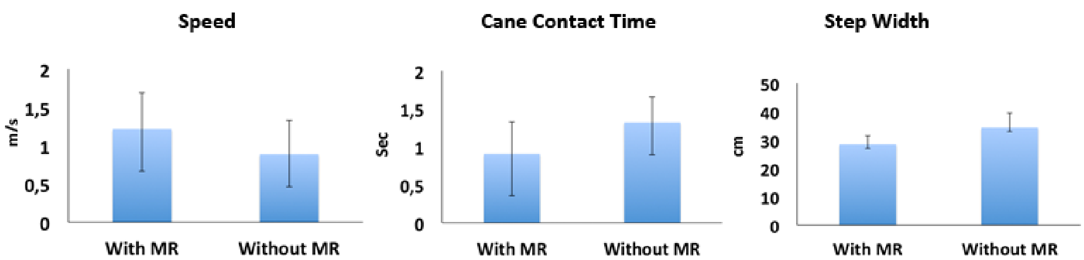

Figure 3. *The relationship between medical recommendations and speed, cane contact time and step width (95% confidence intervals in error bars).*

Pain

The MANOVA found that speed (F (1, 46) = 78.514, p =0.000, n^2=0.636), stance phase (F (1, 46) = 19.475, p =0.000, n^2=0.302), cane contact time with the ground (F (1, 46) = 19.475, p =0.000, n^2=0.704) and maximum elbow flexion angle were all affected by the main effect of pain. Pairwise comparisons showed that those participants who experienced additional pain from using their walking stick had a significantly (p<0,005) lower walking speed (0.74 m/s ±0.28 vs 1.52 m/s ±0.28), higher stance phase (66.81% ±3.2 vs 64.03% ±2.38), higher cane contact time with the ground (0.69 s ±0.18 vs 1.41 s ±0.32) and their elbow angle was more flexed (78.19° ±25.85 vs 102.87 ° ±20.43) than those who didn’t experience any extra pain caused by their walking stick (Figure 4).

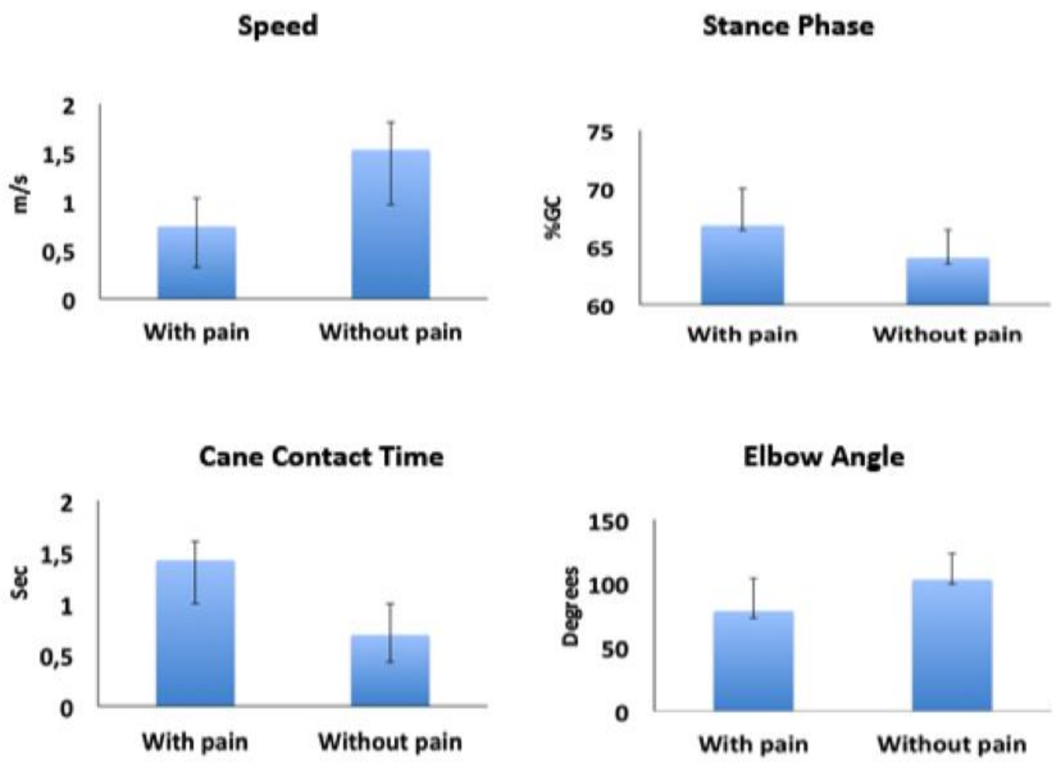

Figure 4. *The relationship between pain and speed, stance phase, cane contact time and maximum elbow flexion angle (95% confidence intervals in error bars).*

Fall History

The MANOVA found that the stance phase F (1, 46) = 53.19, p =0.000, n^2=0.542), step width (F (1, 46) = 20.308, p =0.000, n^2=0.311), cane contact time with the ground (F (1, 46) = 59.135, p =0.000, n^2=0.568), the maximum trunk flexion angle (F (1, 46) = 19.361, p =0.000, n^2=0.289) and the maximum elbow flexion angle (F (1, 46) = 17.297, p =0.000, n^2=0.278) were all affected by the main effect of whether the individual had previously fallen. Pairwise comparisons showed that fallers had a significantly (p<0,005) longer stance phase (68.93±2.89) than the non-fallers (64.67±1.8), took wider steps (34.26 cm ±4.11 vs 28.02 cm ±3.09), their walking stick contact time with the ground was higher than non-fallers (1.38±0.33 vs 0.71±0.24), the trunk angle was more flexed than non-fallers (18.51° ±7.58 vs 10.73 ° ±5.35) and the elbow angle was more flexed than non-fallers (77.07° ±25.37 vs 106.91 ° ±16.23) (Figure 5).

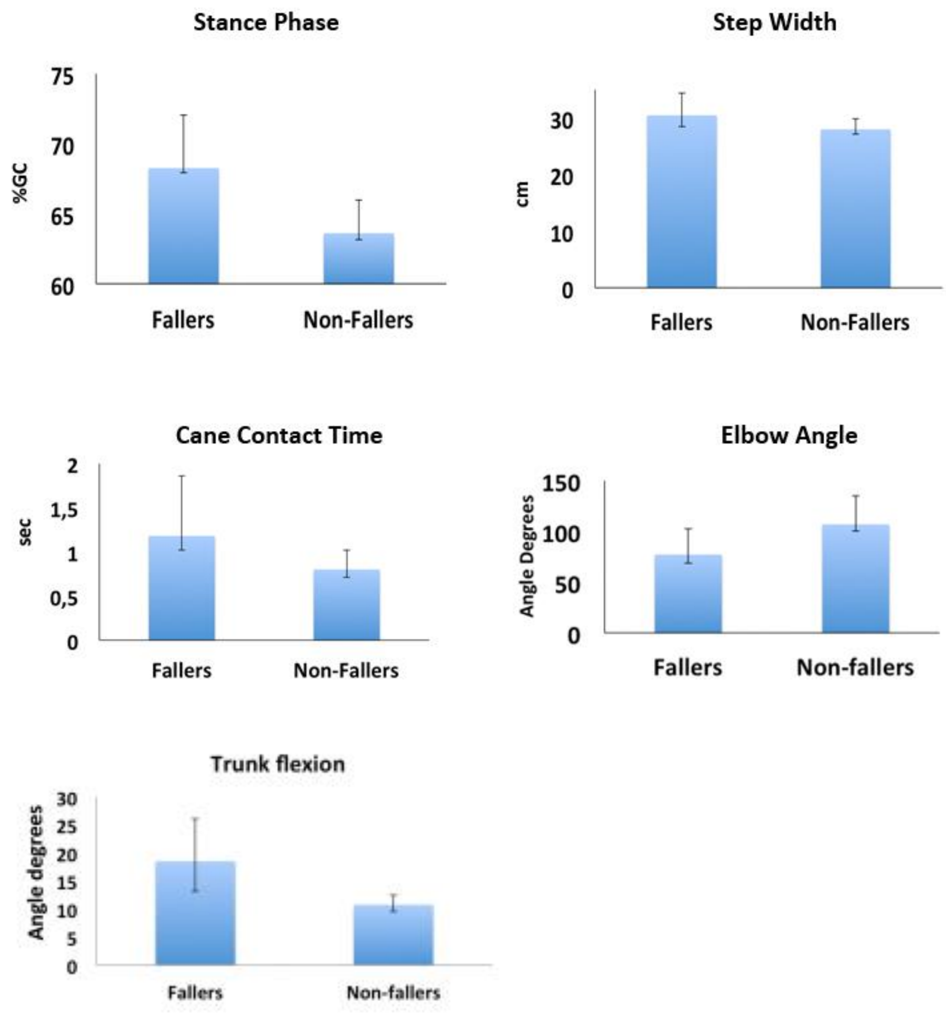

Figure 5. *The relationship between fall history and stance phase, step width, cane contact time with the ground and maximum elbow and trunk flexion angle (95% confidence intervals in error bars).*

3.8 Discussion

The aim of study 2 was to test how the multiple factors identified in the questionnaire affected the gait pattern of elderly walking stick users. The findings showed that speed and step rate decreased with age whilst stance phase and step width increased. Previous studies have suggested that speed can be used as an indicator of walking ability (Salbach et al., 2001; Studenski et al., 2003), which seems to decline with ageing process, particularly after the age of 75 (Branch et al., 2004). In this study we found that the oldest age group (>85 years old) walked more slowly than the other age groups (65-75 years old, 75-85 years old) and took both slower and wider steps that increased the duration of stance phase. This may have been in an attempt to enhance balance and stability during walking. However, this affect was only significant for the >85 age group, not the 75-85 year old age group. The major difference is that our participants used a walking stick whereas Branch et al's (2004) study tested people who didn't use any assistive device. This improvement in gait in the 75 – 85 year old grou[may support the findings of the systematic review by Bateni et al., (2005) that suggested that using a walking stick can improve balance during gait, provide a wider base of support and stability and improve the quality of walking in elderly. This result is important if we take into account that walking ability is related to the individual's perception of a higher quality of life (Schmid et al., 2007).

In this study we also sought to understand whether the medical involvement in prescription and fitting could improve gait performance. Obtaining a prescription involves a medical assessment determining if the person requires a mobility aid and what type of equipment is most appropriate. The assessment is normally carried out at NHS wheelchair services centres or clinics. The people who deliver this assessment are health professionals, such as GPs, occupational therapists, or physiotherapists and they evaluate the individual physical and social needs, as well as the environment in which the patient lives.

We therefore hypothesized that people who have been prescribed a specific walking stick by a GP or orthopaedic specialist would be able to better use it to compensate for their gait disorders and walk better than those who had self-prescribed. Our results support this hypothesis as people who followed a medical prescription for their walking stick use from a specialist were found to walk faster, with a lower cane contact time with the ground and with shorter step widths. These parameters suggest a better gait performance than in those participants who didn't have a medical recommendation. To date, even though previous research highlighted the lack of medical involvement (Bateni et al., 2005; Patcharawan et al., 2014) in walking stick use, there are no published data comparing people with and without recommendations. Therefore, in agreement with Bateni et al., (2005) it can be deduced that a lack of medical consultation and prescription can lead people to abandon their stick or to inefficient use of their walking stick, which can cause pain or increase the risk of falling. Another interpretation might be that people who use a stick without a prescription might not actually need one, and using it makes their gait condition worse.

Pain was also expected to be a factor that decreased gait performance. The results showed that those participants who experienced additional pain from using their walking stick walked slower, with higher stance phase and higher cane contact time with the ground than those who didn't experience any extra pain caused by their walking stick. These results are in agreement with De Kruijf et al., (2015) who found associations between reduced speed, reduced pace, higher stance phase and lower limb pain related to osteoarthritis. In fact, our findings confirm that pain can affect human gait pattern and reduce walking performance. Furthermore, the elbow angle seemed to be different for these two sub-groups. The elbow of those who declared feeling additional pain while using their walking stick was in flexion (78.19°). In contrast for those who declared that their walking stick didn't cause any extra pain the elbow angle was in extension (102.87°). Pain measured in several locations and pain interference with daily activities was associated with greater risk of falls in the elderly (Barclay, 2009). Understanding the combination of factors leading to falls is still a challenge for research.

Finally, we investigated how the history of falls as recorded in the questionnaire (study1) affected the gait pattern of elderly walking stick users. Our findings showed that fallers walked slower, took wider steps with higher stance phase and higher cane contact time with the ground than non-fallers, presumably in an attempt to find better stability. Trunk angle was also found to be more flexed in fallers (18.51°) compared to non-fallers (10.73°), which indicates that body weight was shifted forward in participants with a recent fall in their history. This result was in agreement with Liu et al., (2011) who suggested that elderly individuals with a more forward-leaning posture had a higher rate of falls than those walking with a more upright posture. Furthermore, the elbow for fallers was in flexion (77.07°) whilst for non-fallers it was in extension (106.91°). Interestingly the contact time of the walking stick with the ground and the contrast in elbow flexion-extension angle follow the same pattern for fallers as for those who experienced pain whilst using their walking stick. These findings supported the hypothesis of this study, that having had a previous fall in the last 12 months can affect the gait pattern and reduce walking ability.

3.9 Conclusion

The findings of these studies provided descriptive, kinematics data and related information about the use of walking sticks in elderly populations with moderate gait impairments. In sum we found a lack of medical consultation in the prescription and fitting of walking sticks. Advanced age, having had previous falls and feeling pain in the joints were factors associated with a slower walking pace with reduction of walking speed, higher stance phase and higher cane contact time with the ground. More research is needed to investigate further physiological, environmental factors related to walking stick use by the elderly.

CHAPTER 4: BIOMECHANICS OF ASSISTED LOCOMOTION IN THE ELDERLY: A COMPARISON OF LEVEL WALKING IN A LABORATORY VS FAMILIAR OUTDOOR ENVIRONMENTS.

4. Biomechanics of assisted locomotion in the elderly: a comparison of level walking in a laboratory vs familiar outdoor environments

Study design (Maria Ntolopoulou, Susannah Thorpe); experimental design, ethics and data collection (Maria Ntolopoulou); data analysis (Maria Ntolopoulou), data interpretation and chapter writing (Maria Ntolopoulou, Susannah Thorpe).

4.1 Introduction

Level walking is required for our movement and interaction within various environments and it is an essential daily activity (Dicharry, 2010; Wren et al., 2011). Locomotor patterns of level walking change through the lifespan and balance is reduced in the elderly (Winter et al., 1990). Numerous studies in the lab have measured the parameters of normal and pathological gaits in participants from across the lifespan (Saunders et al., 1953; Perry et al., 1992; Grabli et al., 2012). Nevertheless, examining level walking in outdoor environments constitutes a significant gap in the literature (Allen et al., 2015) because lab environments are unlikely to replicate the environmental complexity of outdoor conditions. In fact, outdoor walking, especially in an unknown environment (Tinetti et al., 2010; Panel on Prevention of Falls in Older Persons AGS, 2011; Thaler-Kall et al., 2015) appears to be more challenging than indoor walking as the reported number of outdoor fall accidents in the elderly was approximately 50% higher than the number of fall accidents indoors (Li et al., 2006; Manty et al., 2009). More than one-third of elderly residents of nursing homes fall at least once a year (CDC, 2006; Tinetti, 2014) and women are more likely to experience severe injuries such as fractures (Tinetti et al., 1995). Todd et al. (2007) suggests that falls are multifactorial and that there are complex relationships between the surrounding environment and the number of falls. Previous research has demonstrated that the risk of falling is determined by intrinsic factors such as age, gender, health status and mobility impairments, and extrinsic factors related to the interaction with the environment such as lighting conditions, slippery ground or uneven surface, clutter, shoes, clothes or inappropriate walking aids (Todd and Skelton, 2004). Therefore, performing the same gait analysis but changing the contextual (shoes, clothes, walking surface, gradient) or the environmental (indoor, outdoor) constraints could potentially cause alterations in gait kinematics and spatiotemporal parameters (Han, 2005). Current literature of level walking kinematics in the elderly is based on clinical evaluations lacking ecological validity (Kaye et al., 2012). A more representative and multifactorial assessment is required in order to develop mechanisms and new, individualised interventions regarding older adults' ability to take part in outdoor activities (Allet et al., 2008). In addition, to date there is no evidence about how the more vulnerable elderly populations such as walking stick users and osteoarthritis patients walk on a flat surface outdoors. It has been suggested that environmental modifications can prevent falls especially in high risk groups such as multi-fallers or elderly with vision impairments (Cummings et al., 1985). However, further research is needed to describe the implications of environmental modifications in gait and the risk of falling, especially in vulnerable groups. Therefore, in this study we investigated environmental constraints, comparing lab and outdoor walking in elderly walking stick users.

Josephson et al. (1991) were the first to suggest that the environment can be a contributing factor in most falls in the elderly. Carter et al. (1997) studied home hazards such as cluttered rooms and slippery surfaces, concluding that in 80% of homes inspected, the elderly lived in a potentially hazardous environment. Roger et al. (2001) in a qualitative study about hazard recognition, highlighted the fact that the elderly may not always be aware of the increased risk they are running and do not use any assistive device to support balance and increase their safety. A large proportion of people older than 65 years are prone to falls due to frailty and instability during level walking, decreasing the area of confidence which is perceived as a safe place to move (Tuunainen et al., 2014). Therefore, simple tasks requiring movement outdoors are perceived as a risk, as they are outside the area of confidence. Mobility aids such as walking sticks can increase safety by compensating fatigue and muscle weakness caused by musculoskeletal pain and diseases (Maki et al., 2005). It is estimated that 24% of the population over 65 uses a walking stick (Resnik et al., 2009). Furthermore, in a residential care setting there are routes accessible to residents such as lighted non slippery corridors or ramps with safety rails that can increase mobility, allowing easy movement in the environment and reducing the risk of trip hazards and confusion (Carr, 2011). When gait is impaired, the movement in the environment is not an automatic motor task but more complex, requiring more attention (Tombu et al., 2003). In fact, walking in a challenging environment (unknown location, irregular surface, clutter) can increase the demands for attention, meaning that more visual focus and sensory input information is needed (Yogev-Seligmann et al., 2008). Therefore, the elderly who experience gait disorders would need extra support for walking in a challenging environment. Since walking sticks provide balance, stability and support during walking, they can increase the area of confidence, the independence and mobility of elderly walking stick users (Van Hook, 2003; Constantinescu, 2007).

Mobility in the elderly can be assessed with kinematics-based gait analysis (Bridenbaugh, 2011). The analysis of walking parameters can be used as a means of evaluation of both well-being (Hodgins et al. 2008; Abellan et al., 2009) and the risk of falling (Verghese., et al 2009) in the elderly, as step length, width, rate and variability are potential indicators of health status (Brach et al., 2007; Studenski et al., 2011). An impaired gait can indicate a physiological and general health decline, increasing the risk of falling. Much of the current literature suggests that walking speed and step length are key parameters in indicating whether the risk of fall is high or not (Maki et al., 1997; Abellan van Kan et al., 2009; Callisaya et al., 2012). Previous research has generally shown that, because of age-related neuromuscular changes, elderly people tend to walk slower than young adults, largely as a result of a reduction in step length (Laufer et al., 2005; Kurz and Stergiou, 2003; Wall et al., 1991; Winter et al., 1990). It has not been established yet whether these reductions are only due to the physiological ageing process or apprehension of a fall accident, or due to both (Chamberlin et al., 2005). Furthermore, it has not been concluded yet whether these alterations in gait parameters can lead to an increase of safety and stability for the elderly. In fact, a slower walking pattern has been assigned as a factor in increasing falls (Cromwell et al., 2004; Ness et al., 2003) and instability (Coppin et al., 2006).

The slower gait pattern in the elderly is also characterized by a reduced step rate that causes an increase of gait cycle time and therefore an increased stance phase and double support time (Riley et al., 2001). This slower pattern and especially this speed reduction was found particularly present in the elderly who have been affected by mild mobility impairments such as osteoarthritis (OA) and they also exhibited substantially higher variability in gait parameters compared to young, healthy participants with no gait impairment (Gu Kang, 2007; Dingwell and Cavanagh, 2001). Increased variability of gait kinematics and spatiotemporal gait parameters has also been associated with an increased fall risk (Maki, 1997) as it was also found higher in elderly fallers compared to elderly non-fallers (Hausdorff et al., 2001; Brach et al., 2001; Brach et al., 2007). Variability can be described as the increased randomness and noise in the musculoskeletal system (Kurz and Stergiou, 2003). The magnitude of variability can be calculated with metrics such as the coefficient of variation (CV=100%*SD/mean) and can be a predicting variable for falls (Hausdorff et al., 2001). Thus, the variability of kinematics and spatiotemporal gait parameters can be used as a tool to characterize locomotor patterns and assess instability during walking (Toebes et al., 2012). Data from several sources have identified changes in gait parameters when the gait analysis was performed in different environments. The variables showing difficulty of adaptation in a different environment or on an irregular surface were decreased speed (Rogers et al., 2008), increased step variability associated with a history of falls (Richardson et al., 2005) or injuries caused after a fall accident (DeMott et al., 2007). In addition, it has conclusively been shown that the range of motion in joint angles of the trunk and lower limbs were higher in order to facilitate forward movement and motor control while exploring a challenging surface (MacLellan et al., 2006; Marigold et al., 2005).

Previous literature has also suggested that in order to make a qualitative distinction among gait patterns (i.e. indoor and outdoor), measurements such as horizontal head acceleration can detect variations in gait motion control that speed and simple gait metrics cannot (Lowry et al., 2013). A number of studies report that controlling head movements during gait could stabilise the visual field, improve the process of the vestibular system signals and ensure better balance. In normal gait, the trunk attenuates head linear acceleration; however, when gait is impaired, this motor behaviour is altered and attenuation is reduced (Mazza et al., 2008; Holt et al., 1999; Berthoz et al., 1994; Cappozzo et al., 1981). In order to prevent falls in elderly walking stick users and reduce risk factors, it is important to first understand the biomechanics of level walking and their adaptations to different environmental conditions.

Therefore, the aim of this study is to compare the kinematics of elderly walking stick users for indoor and outdoor level walking conditions. More specifically, I sought to identify parameters that can be important for gait stability in assisted locomotion and how they are influenced by the environmental effect of lab or real outdoor conditions. It was hypothesized that walking outdoors on a pavement would be more challenging than walking on a flat surface in the lab. It was expected that the lab's flat non-slippery surface would be perceived as a safer and more controlled environment for the subjects even though the pavement near their nursing home is considered their natural environment. Thus, greater speed, step length and step rate were expected in indoor rather than outdoor walking. As the outdoor environment is more extended, more cluttered, and complex (with longer paths), we expected that it would be more difficult for the participants to attend to all the necessary variables and that they would therefore focus their attention more on their path to look ahead more, expressed by higher trunk and neck flexion during walking.

4.2 Method

4.2.1 Participants

Eighteen elderly walking stick users (11 female 83 years ± 7.4, 7 male, 79 years ± 8.7) were recruited from seven nursing homes in Birmingham, UK. All participants were affected by osteoarthritis but were able to walk 150 m. They all used a walking stick in their daily routine for indoor and outdoor activities. The experimental procedure was approved by the ethics committee at the University of Birmingham.

4.2.2 Task and procedure

The experiment was conducted in two steps. Initially participants who volunteered to take part in this experiment were filmed while they walked with their own stick on level ground outside their nursing home in an open place where a 10m walk was available. The selected surfaces were all outdoor flat pavements. Five gait cycles were recorded from the side and face on with a 25Hz Video Camera, and kinematics were analysed using the Tracker 4.90 video analysis and modeling tool. Then participants were asked to walk with their own stick over a 10m walk on a wooden flat surface in the Kinesiology laboratory at the University of Birmingham. The wooden walkway had a safety rail adjacent to it, although no participant used it. The kinematic measurements of movement were captured by 13 infrared cameras of the Vicon Nexus System (Oxford, UK) at a sampling rate of 250 Hz and the Plug-in-Gait was used as the biomechanical model. The thirty-six reflective markers placed on the full body (Figure 1) included the 7th cervical vertebrae, sternum, shoulders, elbow, anterior-superior iliac spine, knee, ankle, heel.

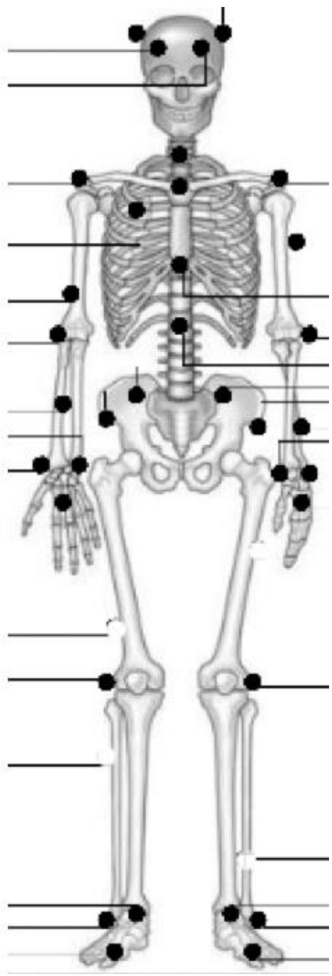

Figure 1. *Marker placement on the full body*

A headband with 4 markers and a wristband with 3 markers were also used. The torso and pelvis markers were applied over a specific uniform with which we provided participants. The markers of the shoulders and legs were directly applied on the skin using hypoallergenic double-sided tape. Each participant performed 10 trials for both walking conditions and five gait cycles were recorded. The maximum joint angles of the ankle (plantarflexion), knee, hip, trunk and neck flexion were calculated by finding the greatest value recorded during the gait cycle across ten consecutive steps in each trial of every participant. The horizontal head acceleration was calculated as the derivate of horizontal head speed found from the horizontal head marker position.

4.2.3 Statistics

Multivariate analysis of variance (MANOVA) was used to test the effects of gender, and the environment condition (lab or outside) on spatiotemporal parameters and kinematics (maximum joint angles) of level walking in SPSS v. 22. The significance threshold was set to p=0.05 after Greenhouse-Geisser correction in order to use a strict criterion of statistical significance.

4.3 Results

For the gait spatiotemporal parameters (Figure 2) the MANOVA results showed that absolute speed, horizontal acceleration of the head and cane contact time were all mainly affected by environmental condition (Table 1) whereas gender was not found to influence the results. Thus, the participant's spontaneous speed was significantly slower outside than in the lab (mean values: 0.74 m/s vs1.22 m/s); the horizontal acceleration of their head was significantly lower outside than in the lab (mean values: 0.49 m/s^2 vs 0.64 m/s^2) and they contacted the ground with their walking stick for longer when they walked outside rather than in the lab (mean values: 0.55 s vs 1.22 s). Figure 2 also demonstrates that step length and step rate were higher in the lab compared to outside and step width was higher outside compared to the lab. However, these differences were not found to be statistically significant.

The only kinematic variable highlighted by the MANOVA tests was neck angle, which was significantly more flexed outside than in the lab (mean values: 21.16° vs 40.61°), but was not influenced by gender. The trunk angle was more flexed for outside walking conditions than in the lab (mean values: 25.14° vs 31.83°). For the lower limbs joint kinematics (Figure 3), the maximum plantar flexion was found lower outside than in the lab (mean values: 76.58° vs 74.23°), and hip flexion was higher outside than in the lab (mean values: 37.18° vs 39.5°), although these differences were not significant.

Tests of Between-Subjects Effects								
Source	Dependent Variable	III Sum of Sq	df	Mean Square	F	Sig.	Partial Eta Sq.
condition	**neck angle**	3402,778	1	3402,778	10,955	**0,002**	0,244
	trunk angle	403,474	1	403,474	2,123	0,154	0,059
	elbow angle	999,508	1	999,508	1,208	0,279	0,034
	ankle angle	49,585	1	49,585	0,5	0,484	0,014
	hip angle	48,279	1	48,279	0,75	0,392	0,022
	speed	2,059	1	2,059	16,579	**0,0000**	0,328
	horizontal head acceleration	2,157	1	2,157	13,675	**0,0000**	0,367
	cane contact time	3,987	1	3,987	30,708	**0,0000**	0,475
	step length	604,067	1	604,067	1,96	0,171	0,055
	step width	74,544	1	74,544	0,279	0,601	0,008
	step rate	551,31	1	551,31	2,532	0,121	0,069

Table 1. *Summary table of multivariate analysis of variance investigating the effect of environmental conditions (in the lab or outside) on gait kinematics and spatiotemporal parameters of elderly walking stick users. The level of significance was set at p< 0,005 (Error Bars with Standard Error).*

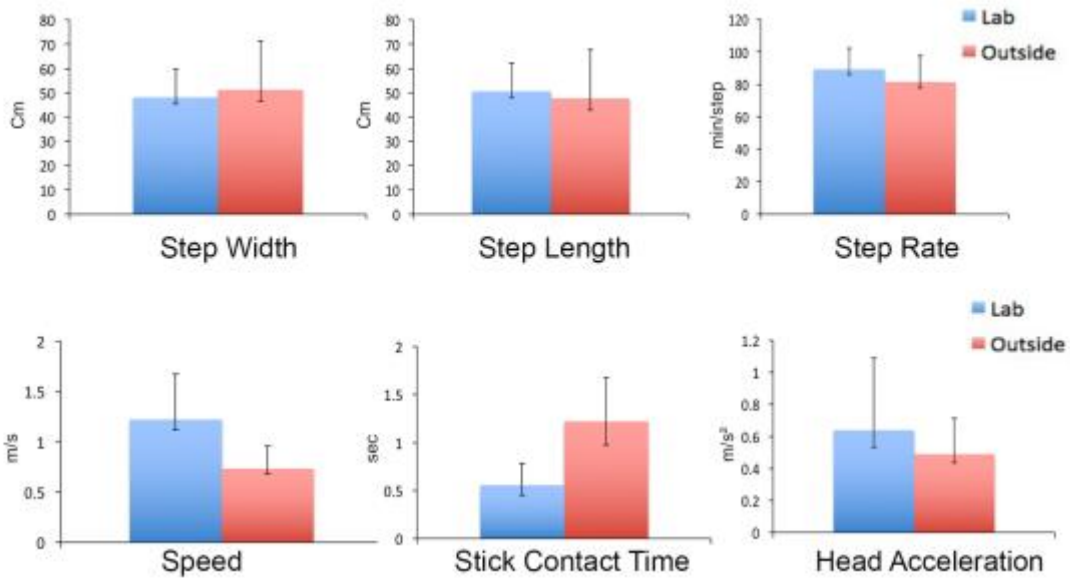

Figure 2. *(a) Spatiotemporal parameters overview: Raw step width (cm), Raw step length (cm), Step rate (min/step), (b) Speed (m/s), Stick contact time (SCT) (s), Horizontal Head acceleration (HA) (m/s²) (95% confidence intervals in error bars).*

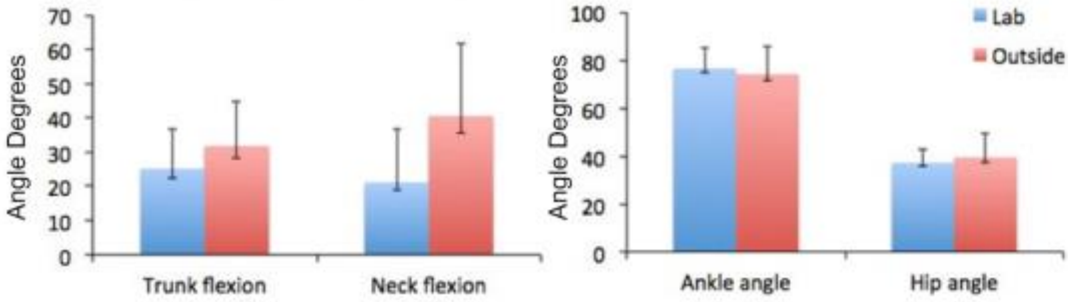

Figure 3 *Upper limbs kinematics: Trunk and neck flexion (angle degrees), (d) Lower limb kinematics: Ankle angle (maximum plantarflexion) and maximum hip flexion (angle degrees) for lab and outside conditions (95% confidence intervals in error bars).*

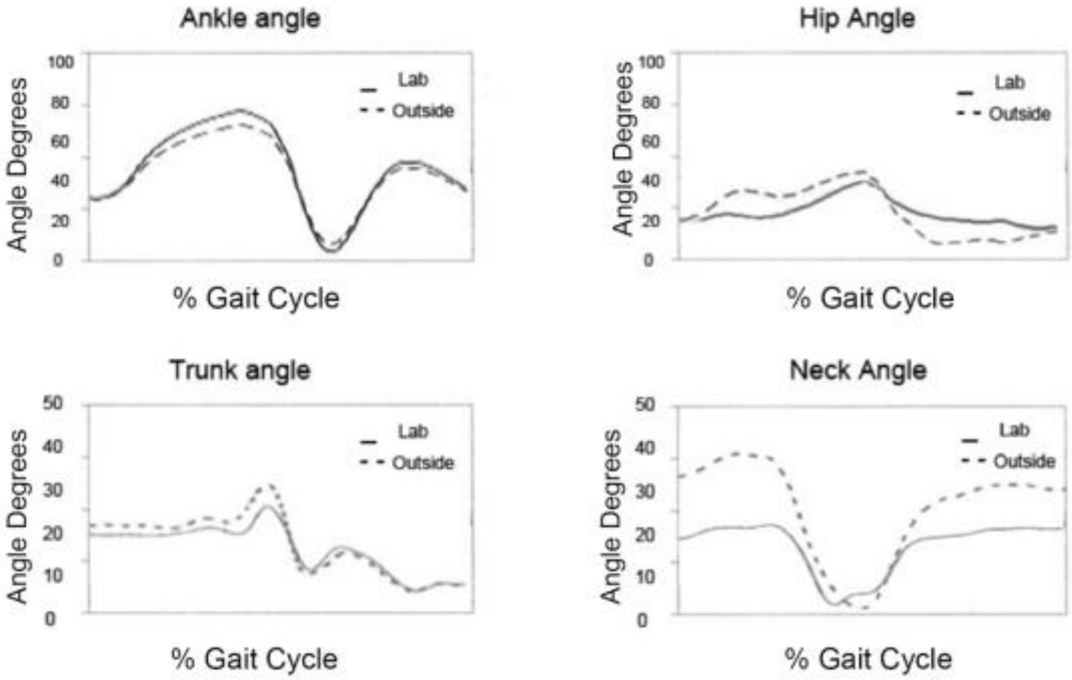

Figure 4. *Ankle, hip, trunk and neck angle profiles illustrating the group average patterns of lab and outside walking for elderly walking stick users.*

The variance and the magnitude of variability were examined by measuring standard deviations and the coefficient of variation of spatiotemporal parameters and kinematics. Both standard deviation and coefficient of variation were found higher outside (Table 2).

	Standard Deviation		Coefficient of Variation	
Gait variables	Lab	Outside	Lab	Outside
neck angle	12,90349	21,32422	8,123916831	15,298362
trunk angle	11,71535	15,58396	46,59733852	48,9488696
elbow angle	27,22606	30,22716	28,23108573	35,1880954
ankle angle	8,33055	11,35713	10,87728003	15,2979765
hip angle	5,83752	9,72783	15,69670144	24,6239525
speed	0,44985	0,21459	28,80438085	36,7724508
cane contact time	0,24951	0,4443	36,34528057	45,7308289
step length	16,29814	18,72624	26,74847635	35,5077078
step width	11,53422	20,04703	22,53658413	41,5034839
step rate	13,11345	16,23492	14,69421821	19,9408053

Table 2. *Summary table of standard deviations and the coefficient of variation of spatiotemporal parameters and kinematics*.

MANOVA found that the coefficient of variation of the step length (mean values: 26.74% vs 35.5%) and step width (mean values: 22.51 % vs 41.5%) variability were significantly lower (p=0.000) for the lab compared to outdoors (Figure 5).

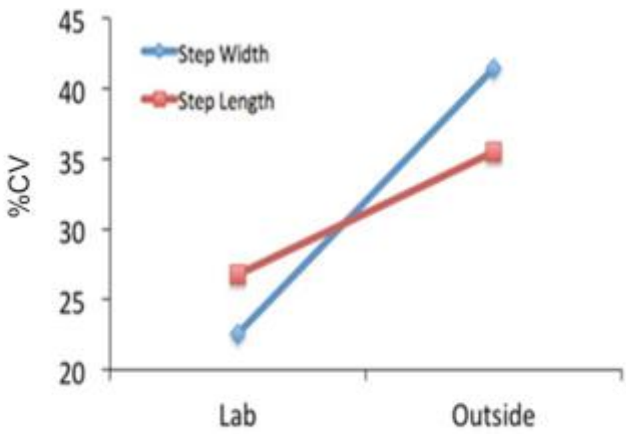

Figure 5. *Step length and width coefficient of variation (CV%) reflecting step variability for lab and outside conditions.*

4.4 Discussion

The first aim of this study was to compare spatio-temporal parameters and kinematics of level walking by elderly walking stick users in indoor and outdoor environments and identify parameters that can be important for gait stability. MANOVA revealed that speed, neck flexion, head acceleration and cane contact time are four key variables that were significantly different when participants performed the same task of level walking in the lab and outside. Participants walked slower, maintained longer contact times between their cane and the floor and exhibited a higher forward flexion of the neck when walking outside compared to inside. Horizontal head acceleration was more attenuated in the lab, head orientation was more forward and subjects kept a more upright position with less trunk flexion. It is necessary to point out that speed was found significantly different between the lab and outside conditions however, its two mathematical constituents (step length and stride duration) were not significantly different between conditions. This is due to the difference between biological and statistical significance. The statistically significant difference indicates only that the difference is unlikely to have occurred by chance. It does not mean that the difference is necessarily large, important, or significant in the common meaning of the word. Human gait and human behaviour in general present great variation expressed with the standard deviation. In this case, the standard deviations of the distributions of step length and stride duration were considerably greater than the difference in the means. The conclusion is that although the biological difference exists for stride length and stride duration, it is unlikely that it is statistically significant.

Overall, a smoother walking pattern was presented in the lab. As suggested in previous literature, outdoor environments are more complex (Tinetti et al., 2010) and therefore, a probable explanation could be that when elderly walking stick users walked outdoors, they required more time to explore. The wooden flat non-slippery surface could also be perceived as a safer and more controlled environment than the pavement in their nursing home, even if it is their familiar environment encountered every day. The bumpiness and roughness of pavements make the surface irregular and potentially perceived as more dangerous than the wooden flat indoor surface and could explain a more cautious and slow pattern in elderly walking stick users. Furthermore, in agreement with Kavanagh et al., (2004) the reduced ability to attenuate horizontal head acceleration may be interpreted as the origins of a rigid and cautious gait pattern. In addition, our findings showed a higher variability when walking in the real world than in the lab for all the tested variables. In agreement with Brach et al., (2001) step width variability was higher than step length variability and both significantly increased when walking outdoors. This result is in congruence with the findings of Richardson et al., (2008) who suggested that walking in a complex environment with a more irregular surface increased gait variability, instability and the risk of fall. Although variability is higher when walking in real environments it is difficult to quantify whether this amount is driven by the complexity of the terrain or by the severity of a pathological disease. The complexity might occur from the fact that in every step, feet have to collect different somatosensory information from a more variable surface and therefore there is an increased effort to find balance, place safely the foot and focus on the path to look ahead. The interaction of these elements with the arthritis pain in the joints might also make the joint movement more variable. Generally, ageing is a multifactorial complex process and the effects causing variations on gait parameters cannot be separated or examined independently the one from the other.

The second aim of this study was to examine the extent to which the level walking condition simulated in the lab reflected real life conditions. In summary, gait analysis in the lab revealed a more stable walking pattern with less variability in all the tested variables than outside. This fact can suggest that gait ability of walking stick users is different in real world conditions and some important key variables indicating instability and risk of falling may be masked when tested only in lab safe conditions. Our results can imply that using only clinical lab gait analysis methods can be a useful proxy for walking in real life for the elderly; however, performance in real environmental conditions should be taken into account even though the measurement tools (video camera) are not as accurate as the tools used in clinical gait analysis (Vicon, force plates, instrumented treadmills etc.) because of the increased variability of gait in natural habitats.

Finally, it should be mentioned that the main limitation of this study was that data collection was performed with two different systems. Previous studies have quantified the difference in accuracy between the two kinds of measures and found that combining Infrared cameras (Vicon) with an alternative system such as a 25Hz video camera (capturing slow movements in low frequency) is a reliable method (Tsushima et al., 2003; Peploe et al., 2014). In our study, the lab gait analysis was performed with the extremely accurate Vicon system, whereas for the outdoor tests simple video cameras (25Hz) were used and subsequently, kinematics were calculated on the Tracker 4.90 video analysis and modeling tool. In addition, as a number of different nursing homes were used, the experimental scene was not the same for all the participants in the outside condition. Nevertheless, we consider it important that this fact could increase variability, as the collected data were ecologically valid. In addition, our analysis included the within subject variability that renders the results relevant.

4.5 Conclusion

The results demonstrated that level walking in the lab showed variations when compared to outdoor walking. A smoother walking pattern was found in the lab with higher speed, shorter contact times between their cane and the floor, and a more upright body posture during walking. Contrary to this, walking on an outdoor pavement increased forward lean, gait variability, and slowed the walking pace, all of which can be an indicator of increased risk of falling. The ecologically valid results of this study are deemed important as they reflect the performance of real world conditions and go further than clinical gait analysis.

CHAPTER 5: LOCOMOTION ON SLOPES: HEALTHY YOUNG AND OLDER ADULTS COMPARED WITH WALKING STICK USERS

5. Locomotion on slopes: Healthy young and older adults compared with elderly walking stick users

Study design (Maria Ntolopoulou, Naomi Mountford [MSci Student], Susannah Thorpe); experimental design, ethics and data collection (Maria Ntolopoulou, Naomi Mountford); data analysis (Maria Ntolopoulou), data interpretation and chapter writing (Maria Ntolopoulou, Susannah Thorpe).

5.1 Introduction

It is important to understand gait variation on differing gradient surfaces because in the elderly population more than 50% of accidents are attributed to falls resulting from walking on stairs and slopes (CDC, 2007). Previous studies have shown that slope and stair walking are ergonomically more demanding than level walking in terms of anterior–posterior stability (McFadyen and Winter, 1988) and lower limb kinematics (Prentice et al., 2004) as the foot must move in both the vertical and horizontal plan and generate greater joint moments in the lower body extremities (Reeves et al., 2008). Therefore, according to Berg et al. (1997), walking on slopes is a determining factor of falls in older adults, and is related to chronic musculoskeletal problems (Schwameder et al, 1999) and joint ailments (Kuster et al., 1995, Redfern & DiPasquale, 1997). Falls on slopes can be caused by poor adaptation to the environment of the sensorimotor systems that regulate body balance (ie proprioception, vision or musculoskeletal deficiencies) (Sturnieks et al., 2008). Walking on slopes or negotiating stairs has been shown to be different from level walking in three parameters: anterior–posterior (AP) stability (McFadyen et al., 1988), medial–lateral stability (Gottschall et al., 2011), and toe off (Prentice et al., 2004). Slope and stair walking also differ from each other. The stance phase was longer in slope walking compared to stairs showing that more propulsion was required to go uphill and more AP stability was required to go downhill (Gottschall et al., 2012). Furthermore, the greater stance phase time is required for the foot placement on the next step (Gottschall et al., 2012). Finally, the foot strike is different between slope and stair walking not only because on stairs there is a toe-strike pattern, but also because participants can self-select step length on slopes but not on stairs (Gottschall et al., 2012). Studies by Young-Mi et al., (2013) and Sheehan et al., (2011) suggest that among flat and stairs or sloped surfaces, the risk of falls while walking on slopes was higher than when negotiating stairs or level walking (Christina & Cavanagh, 2002) as moving on a slope to go uphill or downhill requires a coordinated alternation of the right and the left limbs, more propulsion and balance that individuals with impaired motor control might find difficult to implement (Dixon et al., 2010). It is a fact that today ramps for the disabled and slopes are often encountered by populations who are more prone to falls such as the elderly, people with mobility impairments and those who find difficulty in negotiating stairs. The recommended gradient for ramp construction is between 3° to 6° and ramps up to 9° are permitted by legislation (NHS, P&EFE, 2000). However, many natural and existing urban environments such as sidewalks and roadways can often exceed such recommendations. A previous study (Prentice et al., 2004) investigated the locomotor adaptations of limb and trunk kinematics on slope walking, and tested the effect of a range of different inclinations (3°, 6°, 9°, or 12°) on the gait of a small sample of healthy young adults. The results revealed that the trunk flexion increased in all the sloped conditions and it was initiated while participant were still on level ground. The ankle dorsiflexion also increased while walking on both uphill and down slope conditions to support the increased toe clearance, whilst higher knee and hip flexion angles contributed to the body control preparing for the foot strike (Prentice et al., 2004). Based on these findings the aim of my study was to investigate not only the challenges of walking on a range of different slopes compared to walking on flat terrain, but also to compare gait patterns in healthy young and older adults (control groups) and elderly walking stick users.

Findings on healthy adults have demonstrated that walking uphill significantly decreases spatio-temporal parameters such as speed, step length and step rate, whilst ankle dorsiflexion increases (McIntosh et al., 2006, Phan et al., 2013). In downhill walking, speed and step length decrease also more compared to level walking for both healthy and impaired subjects. The maximum ankle, knee and hip angle increase on both uphill and downhill surfaces compared to level ground (Lay et al., 2005). To summarise, uphill and downhill walking require modifications of gait kinematics and provide different challenges for body posture compared to locomotion on level ground. The main challenge of moving uphill is the lack of propulsion whilst for downhill it is the increase of the step width due to the lack of medial lateral stability (Gottschall et al., 2012).

Nevertheless, despite these variations in spatiotemporal parameters, walking on slopes does not appear to be a significant challenge for the healthy adults. In contrast, the gait of elderly individuals is different from healthy adults as because of the weakening of muscles such as the quadriceps femoris that lead to a restriction of range of joint motion in the lower limbs (Miyaguchi et al., 2003; Menz et al., 2007). Gait variations on inclined surfaces have also been found in people with mild impairments such as osteoarthritis (OA) (Mündermann et al., 2004) or stroke (Phan et al., 2013).

Nevertheless in the literature only a few studies have investigated gait on ramps in osteoarthritis patients (Novak et al., 1989; Mündermann et al., 2004 ; Hohee et al., 2013) and to date no study examined how slopes affect elderly osteoarthritis patients using a walking stick to support gait. The transition from walking level on ground to slopes walking influences not only foot kinematics but the entire balance and motion of the body (Jinger et al., 2011). The trunk and the neck flexion angle increase in order to orientate the gaze and search for the spatial information that is necessary to adapt body posture to the slope (Jinger et al., 2011). As previously suggested by Gracovetsky (1988) the trunk position is a regulating point of the gait. An increased trunk flexion can shift the center of mass forward, decrease the step length and therefore cause instability during gait and increase the risk of falling (Kado et al., 2007). Cromwell (2003) observed that when walking on slopes, step length and step width variability increases as the individuals attempt to gain better balance. However, the presence of greater variability, flexed posture, slower speed and shorter steps during gait might be hazardous for the elderly because the combination of these parameters makes them more prone to loss of balance and falls (Leroux et al., 2002; Menz et al., 2003; Leroux et al., 2006; Lee et al., 2006).

Gait is affected strongly by age. Previous studies have widely reported differences in gait between younger and older adults. Valderrabano et al (2007) found a general decrease of physiological capacities in the musculoskeletal system and movement due to ageing. More specifically, gait pattern change as maximum joint motion decrease and pain appears because of joint diseases such as OA (Valderrabano et al., 2007). The reduction of speed, strength and the ability to maintain an upright trunk position is due to the physiological and functional decline of the musculoskeletal system (Doherty, 2003; Menz et al., 2003). In general, the elderly tend to walk more slowly and focus their attention on the path to avoid obstacles, whereas young adults prioritize fast walking, with balanced and efficient movements (Winter et al., 1990; Cromwell & Newton, 2004; Rogers, Cromwell & Grady, 2008). Maki et al. (1997) suggest that a cautious gait pattern, defined by lower walking speed and shortened step length, is adopted by older people to prevent the risk of falling. However, this slow paced walking pattern may be the cause of greater movement variability, loss of balance and ultimately trips and slips (Maki et al., 2005).

Mobility impairments that stem from ageing affect gait. To compensate for them, almost 24% of the population aged above 60 years are thought to use a walking aid as support for better balance (Resnik et al., 2009). Walking sticks can be used to remove weight from injured joints and to prevent overbalancing (Mulley, 1988). However, to date there are no published data on walking stick use by the elderly on uphill and downhill slopes, hence it is not clear how those already affected by a mild gait impairment are affected by the constraint of a non-level surface. There is some evidence to suggest that using a walking stick could improve balance and gait stability on slopes. Jacobson et al., (1997) studied knee and hip joints motion with and without hiking poles on downhill slopes. They found that using hiking poles improved gait stability by reducing loads on both joints. Another study (Knight et al., 1998) found that using a walking stick on uphill and downhill slopes led to a decrease of forward leaning which, in turn, increases respiration efficiency. It is important to mention that the hiking poles are not used in the same way, functionally and mechanically, as walking sticks. The hiking poles are usually used with a light touch providing additional information about the uneven terrain surface and helping the users to maintain a stable walking pace (Wilson et al., 2001). However, the walking stick, in most cases is used to improve mobility (Bennett et al., 1979) by removing weight from the affected joints and assist in maintaining stability during gait (Mulley, 1988).

It is hypothesized that walking performance (expressed in spatio-temporal parameters and joint angle kinematics) will be influenced by the interaction of two factors: the gradient of the slopes and the participant's age. More specifically, other studies have shown that speed, step rate and step length decrease during uphill walking compared to level gait, whereas in downhill walking there could be an increase in these variables because the slope will exacerbate the participants momentum. Therefore we hypothesized that age will have an effect on slope walking. Compared to young adults, the elderly are expected to show lower speed, rate, shorter steps and longer stance phase duration for all slope conditions. Between the elderly control and the elderly walking stick users groups it is expected that walking ability on slopes would be lower for the walking stick users due to their advanced age, the severeness of their impairments and the increased osteoarthritis pain in the joints compared to the elderly control group. This reduction in spatiotemporal parameters and joint range of motion is expected to be more extended on the inclined and declined slopes than to walk on flat surfaces for the elderly walking stick users group increasing potentially the risk of fall (Sheehan and Gottschall, 2012). The second hypothesis is that the use of a walking stick might improve balance by helping stick users to keep an upright posture and potentially improve spatiotemporal parameters. However, we do not expect major improvements in joint kinematics motion. Since osteoarthritis greatly affects gait pattern (even if the use of a walking stick can, theoretically, partly counterbalance certain variables to improve general stability), it is still not clear in the literature whether the stick itself can increase the risk of fall.

Therefore, the key research questions for this study are:

1. What will be the differences in gait kinematics between level and sloped walking in young, older and walking stick users?
2. How can the use of a walking stick improve balance during gait?

5.2 Methods

5.2.1 Participants

Both male and female were recruited. Three experimental groups were formed: young adults (8 female, 7 male; 21 years ± 2; mean height= 173.14 cm ± 7.83), older adults (7 female, 8 male; 76 years ± 8; mean height= 170.81 cm ± 8.85), and elderly walking stick users (8 female, 7 male; 84 years ± 10; mean height= 167.72 cm ± 7.42), comprising a total of 45 participants. One-way ANOVA showed that the elderly walking stick users were significantly older than the elderly control group (p=0.027). The young and older adults were healthy, physically active, between 19 and 93 years of age. Both groups were free from physical impairment that would have prevented them from engaging in walking activity, and there was no clinical history of disease or injury in the lower extremities or of any other disability. Participants from the group of elderly walking stick users were all arthritis patients, able to walk using a walking stick as support for better balance in their everyday activities. The experimental procedure was approved by the ethics committee at the University of Birmingham.

5.2.2 Task and procedure

Participants were asked to walk a distance of 5m on a custom-made wooden ramp (Figure 1) in the kinesiology lab in five conditions. Firstly, the ramp was laid flat on the floor and the participants walked along a level surface. The ramp was then elevated to uphill with a 4° gradient, uphill with a 7° gradient, downhill with a 4° gradient and downhill with a 7° gradient. Ten trials were conducted in each of the five gradient conditions. Each participant performed 50 trials and walked a distance of 250m in total. As the majority of our participants were elderly, we opted for a short distance walkway that recorded at least 3 complete strides. The order in which they completed the gradient conditions was determined by counterbalancing so that an equal number of participants in each group started with each gradient and condition to remove any potential effect that the order may have on the results. Before testing began, the height of each participant and their walking stick were taken and two familiarisation trials, not included in the analysis, were given. A safety handhold was provided but none of the participants chose to use it in the experiments. The kinematic measurements of movement were captured by 13 infrared cameras of the Vicon Nexus System (Oxford, UK) at a sampling rate of 250 Hz and the Plug-in-Gait was used as the biomechanical model with thirty-six reflective markers placed on the full body as described in chapter 3.

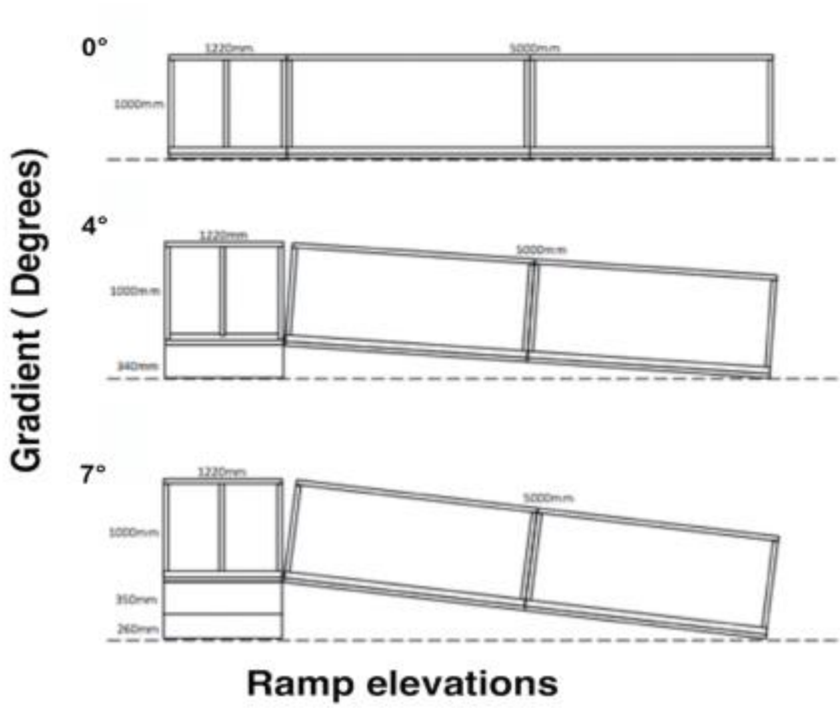

Figure 1. Custom-made wooden ramp elevations at 4 and 7 degrees.

5.2.3 Variables

The participant's gender (female, male), age/ability group (young, elderly control, elderly walking stick users) and the gradient condition (0°, +4°, +7°, -4°, -7°) were defined as independent variables. Ten spatio-temporal parameters of the gait cycle were evaluated. Stance phase duration, walking speed, step rate (steps/min), step length and width were calculated. The stance phase, the time from heel strike to toe off for a single limb, was normalised at 100% so data could be compared within and between subjects. Furthermore, joint angle kinematics such as the maximum ankle plantarflexion, maximum knee flexion, maximum hip flexion, elbow flexion-extension, maximum trunk flexion, neck flexion-extension (as a proxy for gaze focus point) were measured for every gait cycle. The coefficient of variation (CV=100%*SD/mean) was used to define step length and width variability.

5.2.4 Statistics

Multivariate analysis of variance (MANOVA) was used to test the effects of gender (female, male), age/ability group (young, elderly control, elderly walking stick users) and gradient condition (0°, +4°, +7°, -4°, -7°) on spatio-temporal parameters and kinematics in SPSS v. 21. The significance threshold was set to p=0.05 after Greenhouse-Geisser correction in order to use a strict criterion of statistical significance. Where appropriate, Bonferroni post-hoc tests were used to assess differences between and within conditions.

5.3 Results

5.3.1 Spatiotemporal parameters

No gender effect was found to be significant in the analysis. MANOVA found that gait speed and step rate were both influenced by the interaction between age and gradient ($F\ (8,168)= 11.144$ $p=0.00$, $n^2=0.154$); ($F\ (8,168)= 2.494$, $p=0.018$, $n^2=0.154$)). Pairwise comparisons showed that speed was significantly different for all the conditions between all three age/ability groups ($p=0.00$) (Table 1). Figure 2 shows that the lowest speed was found for the elderly walking stick users group who walked slower than the elderly control and young adults. The highest difference was in the -7° downhill condition (0.37 m/s S.D. =0.11) compared to both the level (0.73 m/s S.D. =0.22), ($p = 0.001$) and uphill (0.51 m/s S.D. =0.16) walking conditions ($p = 0.003$). The lowest step rate was found for the elderly walking stick user group, who walked with a markedly slower cadence than the elderly control and young adults. The highest difference was also in the -7° downhill condition (63.74 step/min S.D. =8.16) compared to both the level (78.11 step/min S.D. =6.85), (p = 0.001) and uphill (68.41 step/min S.D. =8.48) walking conditions (p = 0.003). Pairwise comparisons showed that the difference in step rate between young adults and both elderly groups was significant (p=0.00) whereas between the elderly control group and the elderly walking stick user group, no significant difference in step rate was found in the post hoc analysis.

Variables	**Age**		**Gradient**		**Age*Gradient**	
-	F	p	F	p	F	p
speed	164.565	0	127.731	0.00	11.144	0.00
step rate	172.796	0.00	70.152	0.00	2.494	0.018
step width	54.234	0.00	30.058	0.00	1.331	0.24
step length	14.759	0.00	2.928	0.57	0.61	0.762
step length variability	56.481	0.00	31.529	0.00	1.354	0.234
stance phase	92.641	0.00	92.302	0.00	1.642	0.126

Table 1. *Summary table of the significant MANOVA results for spatiotemporal parameters, F (F values), p (p values lower than 0.005).*

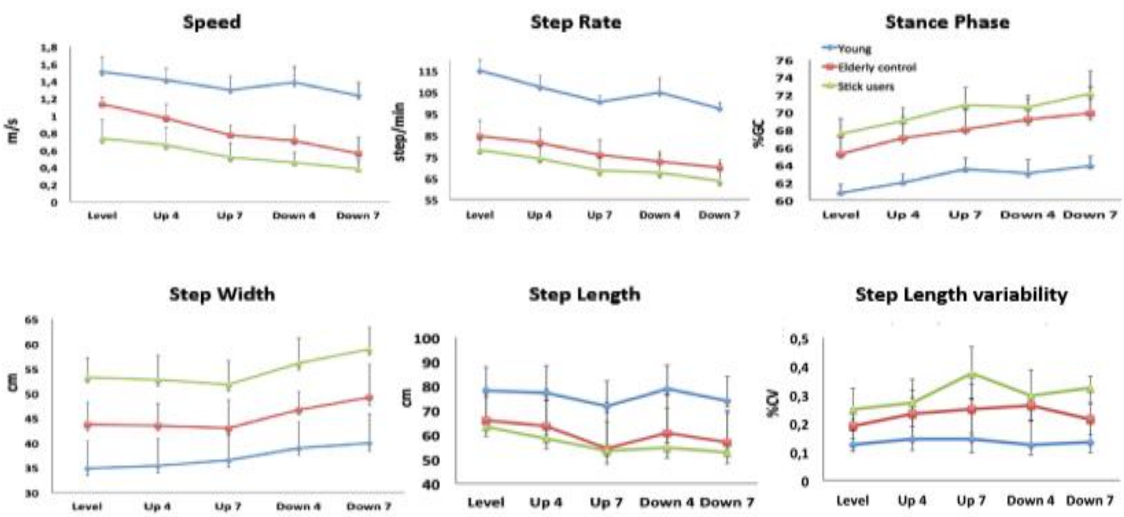

Figure 2. *Gait spatiotemporal parameters for level, +4° and +7° uphill, -4°and -7° downhill conditions for young adults, elderly control and elderly walking stick users. (95% confidence intervals in error bars)*

Step width, step length variability and the duration of the stance phase were influenced by the main effect of age (F (2, 42) = 54.234 p=0.00, n^2=0.291); F (2, 42) = 56.481 p=0.00, n^2=0.318); (F (2, 42) = 92.641 p=0.00, n^2=0.403) and the main effect of gradient (F (4, 168) = 30.058 p=0.00, n^2=0.486); (F (4, 168) = 31.529 p=0.00, n^2=0.413); (F (4, 168) = 92.302 p=0.00, n^2=0.467) but not by the interaction between age and gradient (Table 1). As shown in Figure 2 the step width was significantly different between all the three age/ability groups and the elderly walking stick users took the wider steps than the elderly control and young adults for all the conditions (p=0.00). The step width was significantly higher in the -7° downhill condition (young adults: 39.92 cm S.D. =5.91; elderly control: 49.27 cm S.D. =6.63; elderly walking stick users: 58.89 cm S.D. =4.44) than level (young adults: 34.83 cm S.D. = 5.51; elderly control: 43.74 cm S.D. = 4.5; elderly walking stick users: 53.21 cm S.D. =5.51) walking for all the age/ability groups (p<0.001) (Figure 2).

The step length variability was significantly higher for the elderly walking stick users who took more variable steps than the elderly control and young adults for all the conditions (p<0.001) (Figure 2). The step length variability was also significantly higher in the +7° uphill condition (young adults: 0.14 S.D. = 0.03; elderly control: 0.2 S.D. = 0.01; elderly walking stick users: 0.38 S.D. = 0.09) than level walking (young adults: 0.13 S.D. = 0.02; elderly control: 0.19 S.D. = 0.08; elderly walking stick users: 0.25 S.D. = 0.08) (p<0.001).

Stance phase duration was significantly shorter (p=0.00) for the young adults compared to the two elderly age groups but did not differ between the two elderly groups. The stance phase was found significantly longer in the -7° downhill condition (young adults: 63.87 % S.D. = 1.12; elderly control: 69.89 % S.D. = 2.97; elderly walking stick users: 72.11 % S.D. =2.6), than in level walking (young adults: 60.81 %

S.D. = 1.02; elderly control: 65.29 % S.D. = 1.92; elderly walking stick users: 67.52 % S.D. =1.74) ($p<0.001$).

Step length was influenced only by the main effect of age ($F(2, 42) = 14.759$ $p=0.00$, $n^2=0.273$) (Table 1), and did not differ according to gradient. As shown in Figure 2 the step length was significantly lower for the elderly walking stick users than the young adults. However, between the elderly control group and elderly walking stick users, no significant difference was found in the post hoc analysis. The elderly walking stick users took shorter steps than the two other age/ability groups for all the conditions ($p=0.00$). The lowest step length was found in the -7° downhill condition (young adults: 73.89 cm S.D. =9.9; elderly control: 56.71 cm S.D. =12.14; elderly walking stick users: 52.69 cm S.D. =17.1) and +7° uphill (young adults: 71.71 S.D. = 10.35; elderly control: 54.21 cm S.D. =11.1; elderly walking stick users: 53.28 cm S.D. =20.1) compared to level walking (young adults: 78.01 cm S.D. =9.85; elderly control: 65.77 cm S.D. =12.53; elderly walking stick users: 63.32 cm S.D. =15.78).

5.3.2 Joint Kinematics

MANOVA found that maximum ankle-, knee- and trunk- flexion angles were all influenced by the interaction between age and gradient ($F(8,168)= 2.478$ $p=0.00$, $n^2=0.397$); ($F(8,168)= 3.945$ $p=0.00$, $n^2=0.601$); ($F(8,168)= 3.804$ $p=0.001$, $n^2=0.551$) (Table 2).

Variables	Age		Gradient		Age*Gradient	
-	F	p	F	p	F	p
ankle	42.098	0.00	118.883	0.00	2.478	0.00
knee	32.245	0.00	92.019	0.00	3.945	0.00
hip	2.235	0.12	69.045	0.00	2.509	0.017
trunk	22.035	0.00	36.276	0.00	3.804	0.001
neck	15.737	0.00	19.813	0.00	0.704	0.687

Table 2. *Summary table of the significant MANOVA results for kinematics, F (F values), p (p values lower than 0.005).*

The maximum ankle plantarflexion was lower in both uphill and downhill conditions for all groups compared to level walking. Pairwise comparisons showed that the differences between the three age/ability groups were significant (p=0.00) with the exception of the +4° uphill condition that was significantly different in young adults from the two elderly groups, but not between the elderly control and elderly walking stick users. As shown in Figure 3, the lowest plantarflexion was found in the elderly walking stick user group in the +7° uphill (72.84° S.D. = 9.23) and the -7° downhill condition (58.14° S.D. = 7) compared to level walking (78.23° S.D. = 6.55) ($p < 0.01$).

Knee flexion was higher in both uphill and downhill conditions for all groups compared to level walking. Pairwise comparisons showed that differences in knee (flexion) angle degrees between the three age/ability groups were significant ($p < 0.01$). In the elderly control group, maximum knee flexion was significantly higher ($p < 0.01$) in the +7° uphill (69.79° S.D. = 5.82) and the -7° downhill condition (73.13° S.D. = 5.34) compared to level walking (64.43° S.D. = 4.73) and this emulated the same pattern as for the other age groups. Finally, for the elderly walking stick user group, maximum knee flexion was significantly higher ($p < 0.01$) in the +7° uphill

(62.96° S.D. = 5.01) and the -7° downhill condition (62.37° S.D. = 5.37) compared to level walking (54.69° S.D. = 4.87).

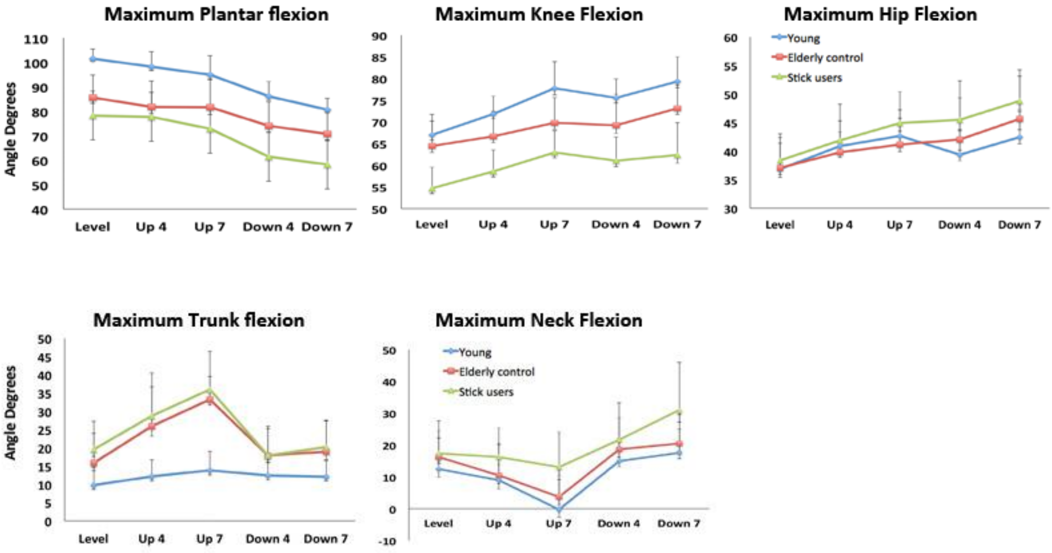

Figure 3. *Lower and upper limb maximum angles of flexion for level, +4° and +7° uphill, -4°and -7° downhill conditions for young adults, elderly control and elderly walking stick users (95% confidence intervals in error bars).*

The trunk lean was significantly lower in young adults than the two elderly groups ($p< 0.01$). However, no significant difference was found in the post hoc analysis between the elderly control group and elderly walking stick users. The trunk lean was found to be higher in uphill and downhill conditions for all groups compared to level walking. As shown in Figure 3, in the elderly control and the elderly walking stick user groups, the forward lean of the trunk was significantly higher for the +7° uphill condition (33.24° S.D. = 6.21, 35.91° S.D. = 10.46) compared to level walking (15.79° S.D. = 8.19, 19.54° S.D. = 7.62). The young adults did not present any significant differences within the slope conditions.

The maximum hip flexion angle was also influenced by the interaction between age and gradient ($F (8,168) = 2.509$ $p=0.017$, $n^2=0.417$) (Figure 3). For the elderly control and the elderly walking stick users' hip flexion was higher in both +7° uphill (41.12° S.D. = 4.64, 44.91° S.D. = 5.51) and -7° downhill condition (45.64° S.D. = 7.45, 48.75° S.D. = 5.47) compared to level walking (37.04° S.D. = 4.36, 38.37° S.D. = 4.65), with elderly walking stick users demonstrating the highest hip flexion compared to the other groups in all conditions. Hip flexion was significantly higher for elderly walking stick users compared to the young adults ($p < 0.01$). However, no significant difference was found in the post hoc analysis between the elderly control group and elderly walking stick users.

Neck angle (maximum flexion) was influenced by the main effect of age ($F (2, 42) = 15.737$ $p=0.00$, $n^2=0.431$) and the main effect of gradient ($F (4, 168) = 19.813$ $p=0.00$, $n^2=0.444$); however, the interaction between age and gradient was not found to be significant (Figure 3). As shown in Figure 3 the neck was significantly more flexed for the elderly walking stick users than the young adults. However, no significant difference was found in the post hoc analysis between the young adult and the elderly control group ($p=0.00$). The neck angle was significantly lower ($p < 0.01$) in the +7° uphill condition (- 0,26° S.D. = 9.44, 3.66° S.D. = 9.11, 13.06° S.D. = 11.04) and higher ($p < 0.01$) in the -7° downhill condition (17.6° S.D. = 7.39, 20.46° S.D. = 9.1, 31° S.D. = 14.94) compared to level walking (12.46° S.D. = 9.76, 16.26° S.D. = 8.19, 17.4° S.D. = 10.14).

5.4 Discussion

The first aim of the study was to investigate how slopes affect gait in the different age/ability groups and whether steeper slopes are more difficult to walk for elderly walking sick users. Based on previous studies that have reported that gait kinematics (spatio-temporal parameters and maximum joint angles) differ between sloped walking and walking on flat terrain we hypothesized that negotiating slopes would be more demanding than level walking especially for those individuals who have the most advanced age and severe gait impairments. A slower paced pattern was expected during uphill and downhill walking compared to level gait for the elderly control and the elderly walking stick users' groups.

The results confirmed the two parts of this hypothesis as speed, step rate and step length were lower for all the age/ability groups when participants walked on inclined surfaces and the lowest values were found in the +7° condition, suggesting that the steeper a surface is, the more difficult will be to walk. In downhill walking the results revealed that elderly control and elderly walking stick users took shorter steps, than in other conditions, which presumably reflects the need to oppose momentum.

The fact that elderly walking stick users were more cautious than the other groups when walking downhill and uphill can be also deduced if we take into account maximum neck flexion found in the steeper (-7° & +7°) conditions, suggesting that their attention was focused on the path whereas the young adults demonstrated a more upright head position in these conditions. The high step length variability in the +7° condition could also indicate that walking on inclined surfaces is less stable, constantly modulated in an attempt to find balance and therefore more demanding than level walking. Therefore, slopes have greater effect on participants with advanced age and reduced locomotor ability than younger, more able groups, and this affect increases with the steepness of slopes. These findings are consistent with Prentice et al. (2004) who point out that sloped walking is more demanding and complex in terms of lower limb kinematics in young adults. Extending their study in the elderly and elderly walking stick users, our results lead to the suggestion that ramps of 7 degrees may be too challenging for vulnerable individuals such as elderly walking stick users, to access in both uphill and downhill direction. In fact, the highest variations in all the spatiotemporal parameters and kinematics were found in the steeper uphill and downhill conditions (-7° & +7°) whereas a 4 degree slopes were closer to flat walking. Therefore, a 4 degree angle could be used for ramps to facilitate access to walking stick users.

Kinematic results also suggest that sloped walking can more demanding in terms of motion for the elderly walking sick users. At the lower limb joints the knee and hip maximum flexion angles were higher in uphill and downhill compared to level walking. More specifically the maximum knee flexion was higher in uphill walking compared to level walking and the greatest peak for all the groups was found in the +7° condition. It was slightly lower for the -4 and higher again for the -7° condition for all the age/ability groups than level walking. A greater range of joint motion is required to negotiate slopes therefore the muscles must provide more power, which may cause problems to the elderly as, according to Valderrabano et al (2007), there is a general decrease of physiological capabilities, strength and joint motion with age.

Contrary to the result that knee and hip flexion were higher in sloped compared to level walking, the maximum ankle plantar flexion was lower for all the age/ability groups. This decrease shows a reduced mobility of the ankle and a decelerated flat-footed gait pattern as in these sloped conditions speed, step rate and step length were also found lower. The interaction between age and gradient was significant for the plantar flexion, suggesting that the older and the weaker we are the less ankle mobility we will have especially on uphill and downhill compared to level walking.

The trunk angle was also found to be more flexed in uphill compared to level and downhill walking, indicating that body weight is shifted forwards during ascent. In addition, we established head orientation by measuring the maximum neck angle in order to estimate the attention focus point while walking. We found that neck flexion was higher for downhill walking compared to level and uphill conditions. This result is in agreement with Jinger et al., (2011) who suggested that the trunk and the neck flexion angle of young adults increase in order to orientate the gaze and collect spatial information necessary for body adaptation to the slope can indicate that participants focus their attention on their path only while walking downhill. The neck flexion was found lower in uphill walking compared to level walking for the young adults (Jinger et al., 2011). Our results added that older individuals with severe pain or gait impairments demonstrated higher trunk and neck flexion than young adults while walking downhill. In fact the older and weaker the individuals were and the steeper the slope, the more trunk and neck flexion was shown not only in downhill but also in uphill walking compared to level. Our results suggest that the elderly walking stick users' pattern is different from the elderly control and the young adult groups for the +7° and -7°. In these conditions, the elderly walking stick users seem to lower their head, that is, higher neck flexion, most likely in order to focus on their path. This may suggest that walking stick users were more cautious and focused on their path, especially when walking downhill.

The reduction of speed, strength and the inability to maintain the upright trunk position are related to ageing and osteoarthritis. Therefore, this "cautious" motion was more pronounced for elderly walking stick users, most likely stemming from the combination that they are significantly older than the other elderly group and that they have been affected by a mobility impairment (OA). This decreased gait ability was found in all variables for elderly walking stick users, including the highest step variability. The results of spatiotemporal parameters (reduction of speed, step rate, step length and the increase of stance phase) are in agreement with previous literature suggesting the advanced age affects gait; therefore, as we get older we walk slower, with shorter steps with longer stance phases and we increase the time that both feet contact the ground in order to increase our stability (Winter et al., 1990; Maki et al. 1997; Cromwell, 2003). From our results, stability is inferred from the combination of the following variables: high speed, high step rate, high step length and the shorter of stance phase.

The effect of osteoarthritis can be reflected in the results of joint angle kinematics, which demonstrated that the maximum values decreased for the elderly walking stick users compared not only to young but also to the elderly control group. According to Valderrabano et al., (2007) osteoarthritis increases pain and decreases range of motion in the affected joints. Therefore, this might be the reason why they show a significantly lower knee flexion peak compared to the young adult and the elderly control groups. In addition, previous literature on osteoarthritic gait kinematics suggested that the hip extensors are over-solicited and the trunk is flexed in forward direction (Neumann, 2013). Hip angle analysis revealed that the maximum hip flexion was found in the elderly walking stick users in both sloped and level walking and this can be due to the severity of the gait impairment that this group has compared to the elderly control and the young adults group. Furthermore, the maximum hip flexion was higher for uphill walking compared to level for all groups. This result is logical as the increased hip flexion has been related with the compensation of knee flexion (Perry et al., 1992). The increased hip flexion can destabilize the pelvis and cause the loss of balance (Steultjens et al., 2000). Our results show clearly that young adults have a more stable pattern than the elderly and that those who do not use a walking stick seem more stable in level and sloped walking. According to the previous literature the exaggeration of hip flexion and therefore the excessive hip contractures can cause pain and decrease gait stability (Perry et al., 1992).

Previous literature has suggested that osteoarthritis patients tend to obtain a different body position with the centre of gravity shifted in forward direction (Neumann, 2013). Our results are consonant with Neumann (2013) as the trunk angle analysis reveals that the elderly and elderly walking stick users leaned more forward than young adults for all the conditions of level, uphill and downhill walking. The highest values of the neck flexion were found in the elderly walking stick user group that have a more severe gait impairment due to osteoarthritis.

Walking stick use

It was hypothesized that the use of a walking stick could improve balance by helping stick users to keep an upright posture. Analysis of the trunk angle showed that no significant difference was found between the two elderly groups and, especially for the downhill walking condition, their pattern was almost identical. Furthermore, the difference of the step length and the step rate between the young adults and the two elderly age groups was significant. However, for both variables no significant difference was found in the analysis between the elderly control group and elderly walking stick users. Thus the walking stick users seem to maintain their cadence and step length at a similar level with the elderly control even though their speed was significantly lower. This could support the hypothesis that the use of a walking stick provided support to the participants. Even though the elderly walking stick user group presented overall lower gait performance in all the measured variables, it was observed that elderly walking stick users seem to maintain their step length and closely approach normal elderly gaits, especially in the most challenging conditions of +7° and -7° degrees.

Finally, the increased step length variability was found to be significantly different between the two elderly groups in the steeper slope conditions. This can be interpreted as an indicator of instability and higher risk of falling (Maki et al., 1997; Hausdorff et al., 2001) when walking on a challenging surface, as the lack of consistency in foot placement can destabilize the body movement. From the other hand, it has been suggested that the presence of high variability can affect control of movement but not always in a negative way (Brach et al., 2005; Stergiou et al., 2006 Harbourne et al., 2009 Fetters et al., 2010). These studies reported that the increased variability could also contribute in the attempt to reach stability, indicating a potential modulation towards a more stable gait pattern. Therefore, we do not have conclusive evidence to suggest whether the walking stick users are running a higher risk of falling or whether through this higher variability they regulate and adapt their steps to the slopes. However, we can suggest that higher step length variability was found in individuals with advanced age and impaired gait ability due to osteoarthritis compared to healthy young adults. Step width variability was also calculated but wasn’t found significant in the analysis.

5.5 Conclusion

The main results of this study indicate that sloped walking can be more challenging than level walking for the elderly and particularly the osteoarthritis patients walking stick users. Due to their musculoskeletal weakness, locomotion on slopes is more demanding for these groups than level walking as a greater limb flexion and better movement control is required. Furthermore, the steeper slopes are the more challenging is for individuals with advanced age and gait impairments to walk to on. The maximum ankle plantar flexion decreases whilst the knee and hip flexion angles and forward lean increase in sloped compared to level walking. The gait pattern was found to be slower and more instable in uphill and downhill walking as speed, step rate and step length decrease whereas step variability and the stance phase increase.

Age also affected the gait spatiotemporal parameters and kinematics, as the elderly were slower and more instable than the young adults. The fact that the elderly walking stick user group and the elderly control group were not statistically different in most conditions could corroborate our hypothesis that the use of a walking stick can improve arthritis impairments in the elderly.

Currently the ramps up to 9° are permitted by legislation in the UK (NHS, P&EFE, 2000), whereas many existing ramps in various urban environments can often exceed such recommendations. Overall, based on the results of this study it can be advised that ramps up to 4 degrees angle could be used for to facilitate access to walking stick users. However, ramps steeper than 6 degrees angle can be more challenging and consequently increase the risk of fall for the elderly and particularly the people with mild gait impairments, walking stick users.

CHAPTER 6: GENERAL DISCUSSION

6. General Discussion

The aim of this thesis was to study assisted locomotion in elderly osteoarthritis patients. Five experiments were conducted, each focusing on constraints that affect gait, with a view to understanding the multifactorial decline of locomotion in the elderly. Two critical contact points with the ground were identified and investigated, namely, shoes and walking sticks, since by regulating contact of the body with the ground they provide essential sensorimotor input about the environment.

In chapter two I explored kinematic and kinetic variations in gait through the lifespan in natural locomotion, both in barefoot and shod conditions. Recent research has suggested that shoes play a key role in locomotion because they constrain natural motion (Morio et al. 2009). My aim was to describe the extent and nature of the changes in walking pattern induced by shoes in comparison to the natural control situation of barefoot walking. As the effect of age has not, to date, been systematically reported barefoot walking could be the benchmark against which we assess the effect of shoes over the lifespan.

Previous studies have reported differences in gait kinematics and kinetics between barefoot and shod walking (Lythgo et al., 2009; Oeffinger et al., 2009; Franklin et al., 2015). My findings have confirmed 1) that gait mechanics are different between these two conditions; 2) that ankle mobility reduces as we get older (Lieberman 2012) and 3), that the chronic use of shoes can modify the biomechanical function of the forefoot (D'Aout et al., 2009). The last statement is supported by ankle angle joint analysis in children, indicating two distinctly different patterns: a greater range of motion and maximum plantar dorsiflexion mobility while walking barefoot and a considerable decrease in these in shoe walking. The reversal of this phenomenon in the elderly could suggest that the biomechanical function of our foot is altered across the lifespan. However, this study cannot separate effects of age and habitual use of shoes, because I didn’t have age-matched groups of habitually barefoot people. Generally, the elderly showed a lower range of ankle motion than children and young adults. For the elderly, the highest ankle mobility was found when wearing shoes, and the lowest in barefoot pattern.

The findings of the first study show that our body and our movement change with age and that the walking quality in the elderly might be higher when wearing shoes. The reduced ankle mobility can have implications in falls prevention, especially in barefoot walking. We therefore need to consider that it is very common for the elderly to walk barefoot or with slippers that don’t provide stable cushion in their homes (Munro et al., 1999; Menz et al., 2006) and that walking barefoot or with insufficient cushioning could increase the risk of falling in the elderly, especially in indoor environments (Dunne et al., 1993; Menant et al., 2008).

The results of the next chapters bring together information about other constraints that have a direct link with the increase or decrease of the risk of falling. For example, these include medical problems such as arthritis, pain, walking aids, and the environment. Overall, the sum of this work aimed to highlight the multifactorial nature of falls in the elderly population groups.

In the first study I examined the general decline of the ankle mobility due to ageing, focusing on the constraints of footwear. Elderly participants in Chapter 2 were physically active and able to walk independently without the use of a walking stick. In those affected by osteoarthritis, the symptoms of the disease were not severe enough to cause major gait impairments. Bearing out previous literature (Winter et al., 1990), gait analysis through the lifespan revealed a general decline in walking ability with age. Kinematics showed a decrease in joint range of motion, and the analysis of spatiotemporal parameters suggested an increase in step width, a decrease of speed, step length and cadence. These findings are indicative of a more unstable gait pattern in the elderly (with or without footwear). Gait instability leads to functional issues such as injury or even death as a result of falls, which in the elderly are a complex issue related to the multifactorial decline of gait. (Tinetti et al., 1998; CDC, 2005).

The next chapters shed light on further constraints that affect locomotion towards a better understanding of their relationship with falls and falls prevention. In the third chapter I sought to investigate the progression of gait abilities in osteoarthritis patients who, owing to more severe symptoms and impairments, used a walking stick to support gait. I first analysed the reasons and health conditions that led participants to use a walking stick. Based on previous studies suggesting that walking sticks can improve balance, mobility and reduce joint pain (Bateni&Maki, 2005), I investigated how walking sticks had been obtained, fitted and used. Furthermore, the history of patients' falls in the last twelve months was recorded. The findings from the questionnaire revealed that a large number of elderly osteoarthritis patients who suffer from moderate gait impairments decide to use a walking stick to compensate for osteoarthritis pain without medical involvement in the selection, use and fitting of walking sticks. This is consonant with Bateni and Maki (2005) who reported that, of the large proportion of users experiencing difficulties with stick use, 30%-50% abandon their device soon after acquiring it. In addition, we confirmed the lack of evidence available to underlie guidelines on selection and use of an appropriate stick (e.g. which side to hold it on, how high it should be, how far in front of the body patients should strike the ground). In conclusion, it is clear that ineffective use of walking sticks and subsequently, greater risk of falling, can be put down in part to the lack of medical involvement, prescription of a specialist and precise guidelines.

The second study of the chapter was a gait analysis that links the multiple risk factors identified in the questionnaire to the gait pattern of elderly walking stick users in their natural environment, in an attempt to understand how gait parameters are affected by health condition, previous falls and the use of a walking stick. The results showed that those with a history of falls and more severe osteoarthritis pain walked at a slower pace with wider steps. However, the most interesting finding was that a large number of walking stick users reported that using their device for prolonged periods (more than half an hour without a break) caused additional pain, not only in the affected joint but in other joints as well. This suggests the osteoarthritis patients may be using non-affected limbs in a more demanding or different way when they are using their walking stick than when they walk without it: a rigid handle, for instance, or the wrong height of the stick (too short or too high) can eventually affect other joints such as the fingers, hands, elbows or even increase lower back pain due to the forward leaning position of the trunk. The elbow was the variable with the biggest within subject variability, showing that elbow position was different for each step and each contact of the stick with the ground - this may also be explained by increased gait variability or inappropriate fitting of the walking stick or that it is not functionally important. In short, the above may be viewed as an example of how an impaired gait pattern and the inappropriate use of the walking stick can increase osteoarthritis pain. In agreement with Barclay (2009), it can be speculated that pain spread in other locations can increase the risk of falling and decrease the amount of walking. In the video gait analysis I observed a different elbow pattern for people feeling additional pain while using their walking stick: the elbow was clearly more flexed whereas in those who experienced no extra pain it was almost in extension. However, the interpretation of the elbow angle results has complex implications. People who experience severe pain may place the walking stick in a different position relative to the body than those who do not experience pain. The walking stick placement requires further analysis on its relative position according to normalized participant's height. The analysis of the applied forces on the walking stick could also give a better insight on the type of use and the efficiency of gait. Force sensors can be put under the ferrule in order to estimate the applied vertical force on the stick. In addition, the grasping force of the handle could provide a better understanding on whether the participant is using the walking stick to reduce loading from an impaired limb or just for a light touch.

A major implication of chapter three is how the use of a walking stick could prevent falls in the elderly or improve gait and confidence in people who have had a previous fall. Consonant with Rubenstein & Josephson, (2002), the gait analysis results and comparison of gait parameters between fallers and non-fallers demonstrated that suffering a fall in the previous twelve months can affect gait pattern and reduce walking ability. In fact, I found a multifactorial relationship between the severity of health condition, pain, the amount and the quality of walking. People who had not experienced a fall within the last year were found to feel less pain, be more active (including more walking and daily activities), and perform better in gait analysis by demonstrating a higher speed than fallers and shorter contact time of the stick with the ground. In agreement with the previous literature, I confirmed that osteoarthritis is a key intrinsic factor which intensifies the effects of age on gait parameters (Whittle, 2007; Berenbaum et al., 2013) and subsequently triggers interaction with other intrinsic factors such as the fear of falling and the decrease of confidence.

In the following chapters I analysed the extrinsic factors and constraints that affect assisted locomotion in elderly osteoarthritis patients. In agreement with previous studies from Tinetti and colleagues (2010), my results showed that walking in outdoor environments is more demanding than in the lab. My findings imply that the general decline of the visual and vestibular system in the elderly makes this task more difficult. The attentional demands on elderly osteoarthritis patients to extract the necessary environmental information are higher. Furthermore, the neuromotor demands of coordinating their gait with the walking stick and maintaining balance during walking make outdoor walking more challenging and potentially more prone to falls. It is worth noting at this point that the type of surface likely also plays a fundamental role in the production of a cautious gait pattern, as pavements tend to be rougher and more irregular.

Similar characteristics of a cautious gait were also found in the last study, testing locomotion on inclined and declined surfaces. In uphill walking, I found that elderly osteoarthritis patients adopted an increased forward leaning position that shifted the centre of mass in a forward direction, while during downhill walking I found a shift of the centre of mass in the opposite direction. This backward shift, combined with the reduction of speed and step length, can increase the risk of slipping or losing balance (Kado et al., 2007). As suggested by previous studies on individuals without walking sticks, this reduction of gait ability on sloped surfaces may contribute to an increased risk of falling (Kado et al., 2007). I speculate that the risk of falling will be higher for the elderly osteoarthritis patients because of their functional, physiological and psychological vulnerability compared to healthy and active individuals. Nevertheless, according to our results, the reduction of walking ability measured in gait parameters can be attenuated if the elderly osteoarthritis patient uses a walking stick. The results of chapter 5 revealed that the trunk flexion, step length and step rate weren't significantly different between the elderly control group and elderly walking stick users. This could suggest that the use of a walking stick provided support to the participants and allowed them to maintain their walking ability even though they were older and the severity of osteoarthritis was higher for the walking stick users. Therefore, walking on sloped surfaces, especially for elderly osteoarthritis patients, can certainly be more challenging than level walking as it results in a slower, more cautious gait pattern which differs from the participants preferred gait, however the use of a walking stick could reduce the risk of falling.

In addition, it seems that the use of a walking stick can alter bipedalism, producing another type of locomotion, similar to tripedalism. This involves a different type of coordination, weight distribution, timing and symmetry of gait. For this reason, walking stick parameters should neither be neglected nor separated from gait analysis. That means that the walking stick's spatiotemporal characteristics should be included in the analysis and therefore measure variables such as the elbow angle and the stick contact time with the ground.

The use of two walking sticks could provide two extra points of contact with the ground, essentially converting bipedal locomotion into four-legged locomotion. With more points of contact, we could speculate that walking stick users are less likely to slip in the first place, and slips are less likely to turn into falls. The use of two walking sticks could also help the elderly maintain forward posture while walking uphill or on complex terrain, navigate over and around obstacles. Furthermore, engaging both arms to take the load off the knees could be more effective than using just one walking stick alone. This could also prevent joint pain especially in elderly patients with osteoarthritis.

To summarise, my thesis sought to shed light on the relationship between falls, and both intrinsic and extrinsic constraints that affect locomotion in elderly osteoarthritis patients - this relationship includes the physiological decline of ageing, the effect of osteoarthritis disease, the physical and psychological effects of a fall and all the environmental constraints that can decrease the amount of walking, downgrade the quality of patients' lives or may even lead to mortality (Figure 1).

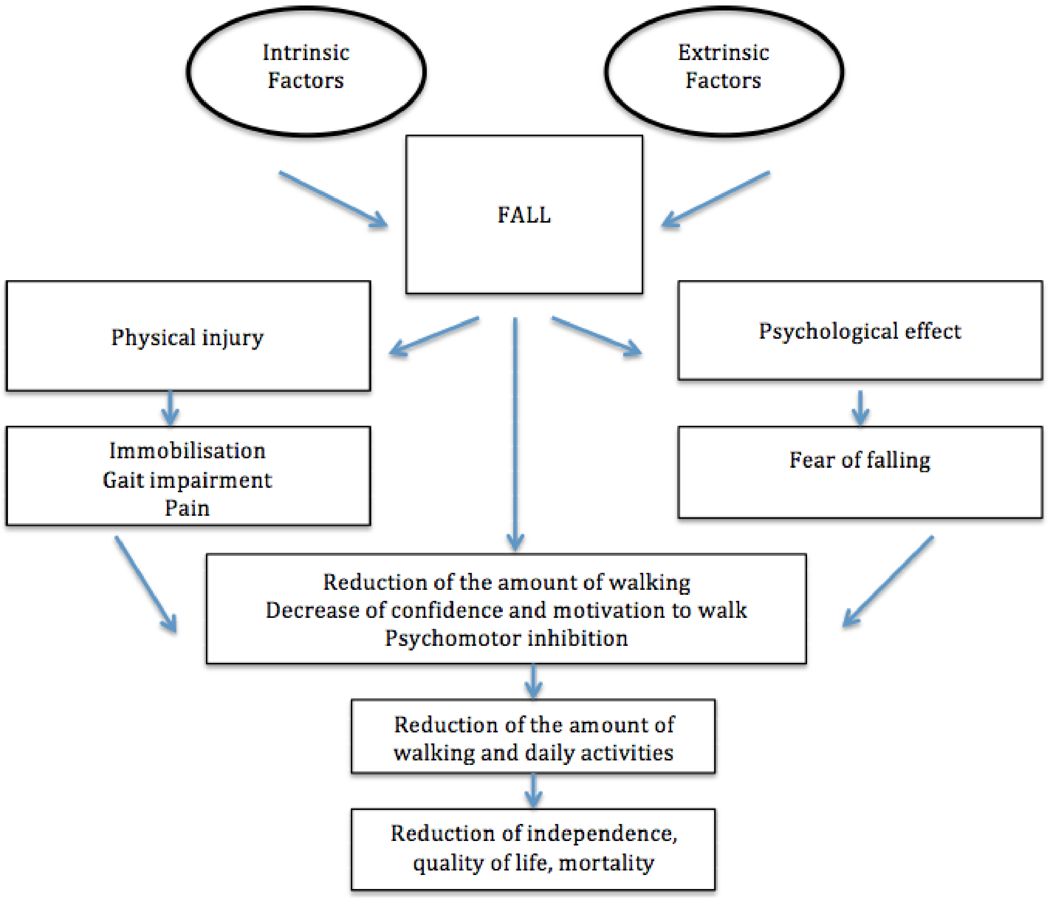

Figure 1. *Relationship between intrinsic, extrinsic constraints and falls affecting locomotion in elderly osteoarthritis patients.*

This link can describe the vicious cycle between intrinsic and extrinsic constraints and falls. It is vital to stress the broader implications of this study and therefore look at the psychological effects caused by a fall: the growing fear of falling reduces the amount of physical activity and increases isolation. The use of a walking stick can become a regulating parameter by potentially improving gait, as shown in chapter 4 where walking stick users were close to the control group even though they were older and osteoarthritis pain was more severe. Conversely, we can deduce that if a walking stick is not efficient due to an inappropriate fit to the users morphology, or causes additional pain, it can potentially increase the difficulty of the walking task and become yet another constraint affecting locomotion. In future research it would be interesting to examine the walking stick height relative to the person's height in order to test the efficiency of the walking stick use.

<u>The major findings of my work regarding walking stick use:</u>

• it can increase stability during the gait of osteoarthritis patients

• it allows osteoarthritis patients to have gait parameters closer to that of a healthy, active elderly person.

<u>The major findings of my work on the extrinsic constraints of locomotion:</u>

• outdoor walking can be more challenging than indoor

• uphill walking is more demanding for elderly osteoarthritis patients walking stick users who adopt an increased forward leaning position

• downhill walking is likely to be the most risky gradient for trips and falls as elderly osteoarthritis patients consistently tend to brake

It is necessary to point out one limitation of this thesis: in two of the studies, data was collected from a simple video camera (25Hz), which is not the most accurate means of analysing kinematics. Nevertheless, the slow speed of my participants would indicate that a sampling rate of 25Hz is adequate to track the major patterns in their gait. As regards the comparison between lab and outdoor walking, data collection was performed with two different systems. Furthermore, apart from the final lab-based study of sloped locomotion where participants were provided with the same clothing so that the markers of the upper body could be applied directly on to the skin, in the rest of the study, participants wore their own clothes. Likewise, in the barefoot study, participants wore their own shoes. However, this limitation could also be viewed as one of the strengths of this research since the data collected were ecologically valid. In the literature, there is a gap in ecological data as all studies on assisted locomotion are based on either clinical gait analysis or on surveys only. Therefore, the combination of the two provides new baseline data in the field. In addition, our analysis includes the within subject variability, which makes the results relevant.

It is also necessary to point out that in ageing research there is a lack of true control groups. In fact, it is practically impossible to eliminate all of the confounding variables and bias when two elderly groups are tested. For instance, in chapter five we compared a young adults control group, an elderly control group with the experimental group of elderly walking stick users. The difference between a control group and an experimental group is that one group is exposed to the conditions of the experiment (i.ie walking stick use) and the other is not. It must be noted that it is not possible to completely control all variables especially when we examine human subjects. There may always be variables at work which, as experimenter we are unaware of or that we cannot quantify. In particular, it is impossible to completely control the ageing diseases, the level of pain, previous injury or the mental world of people taking part in a study.

Other strengths of this work could lie in the fact that the same sample of walking stick users was used in the gait analysis of the questionnaire, the gait analysis video study and in the comparison between ecological and lab walking. A large majority of the above sample (85%) was also tested for the study of slope. This takes on significance because the variability between subjects is controlled - a crucial consideration, especially for elderly osteoarthritis patients who suffer from different health conditions.

This work can have broader applications in research, technology and safety. Since it is generally acknowledged that the recommendations for fitting walking aids are not based on scientific evidence, this work provides interesting data that could be used for further research in the field and inform NHS guidelines. Results from the study of areas such as elbow angle and forward lean during walking, and not just from measuring the distance of wrist in standing position, should be considered in the fitting procedure. The design of a smart stick, that is, the optimisation of a walking stick for use by the elderly - for instance a self-adjustable stick adapting its height based on the individual's gait data when the patient walks on uphill or downhill surfaces - is an interesting prospect.

Conclusion

To summarise, the major outcomes of this work indicate that with ageing humans tend to walk slower, with shorter steps, longer stance phases, and to increase stability, there is an increase in the time that both feet contact the ground. The chronic use of shoes affects foot mechanics, resulting in a more flat-footed gait pattern with a reduced range of motion. The effects of ageing and shoes can be compounded by impairments such as osteoarthritis or other environmental constraints and additionally, walking on slopes or outdoors can potentially undermine stability and increase the risk of falls. However, I found that the use of a walking stick can improve gait parameters, provide better balance and support during gait. With this in mind, it is important for the walking stick to be considered as a tool to enhance the quality of life and prevent falls in elderly osteoarthritis patients. Finally, it is essential for the NHS to establish clear guidelines on the use and the fitting of walking sticks according to the type of impairment and the specific needs of the patients.

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APPENDIX

University of Birmingham School of Biosciences
Research Survey on Assisted Locomotion in the Elderly

Can you please take a few moments to complete this survey about your walking stick use? The information you supply will be analyzed as part of a research project on assisted locomotion in UOB.

General information

Gender

Female

Male

In what age group do you belong to:

Under 50

50 - 65

65 - 70

70 - 80

Over 80

What's your weight?

7,5- 8

8 - 8,5

8,5 - 9

9 - 9,5

9,5 - 10

10 - 10,5

10,5 - 11

11 - 11,5

11,5 - 12

12 - 14

Over 14

Accurate weight in stones (if applicable)

Walking stick use

What was the primary reason that led you to use your walking stick?

Arthritis

Knee replacement

Hip replacement

Balance/ General Support

Fear of falling

Other

In which joint was the arthritis? (If applicable)

Which leg is affected? (Not just arthritis but also replacements & other conditions)

In which hand do you hold your stick?

Opposite to the affected leg☐

On side of the affected leg☐

Dominant☐

Both☐

Who recommended that you use a walking stick?

Personal choice☐

Family/ Friends☐

GP☐

Orthopedic specialist☐

Physiotherapist☐

Other☐

What did they advise you to do?

When did you use it first? (dd/mm/yyyy)

Do you use it every day?

How many hours per day do you use your stick, on the days that you do use it?

Where do you use your stick?

Home☐

Outdoor☐

Other activities (shopping, travelling, garden etc)☐

Other☐

Can you walk without your stick?

Yes☐

Yes, but not more than 50 m (165 ft)☐

Yes, but not more than 100 m (330 ft)☐

No☐

How necessary is for you to use your walking stick?

	1	2	3	4	5
Outdoor					
Home					

Do you use any other devices to help you walk?

No☐

Walking stick with 3 legs☐

Rollator☐

Zimmer frame☐

Other☐

Clinical History

Have you had any previous surgery on your legs?

Yes

No

Have you had fall?

No

Yes 1

Yes > 1

Did you have or are you currently following a physiotherapy or rehabilitation program?

Yes

No

What did/do you have to do in that program? (If applicable)

Walking Stick Fitting:

May I measure your walking stick?

Height: Handle angle: Handle type: Weight:

Who fitted the stick for you?

Yourself

GP

Orthopedic Specialist

Physiotherapist

Shop

Manufacturer

Why did you select this particular walking stick?

Liked the design☐

Liked the shape of the handle☐

Liked the adjustable height☐

It is good on flat and bumpy ground☐

The shop manufacturer advised it☐

Other☐

Is your walking stick comfortable?

Yes

No

Does your stick cause you any pain after long use ?

Yes

No

Is there anything that you would like to improve on your walking stick and, if so, what needs improving and how would it be improved?

Would you be interested in receiving information about future work on the optimization of walking sticks and potentially taking part in future experiments?

Yes

No

Contact details: