Investigating gait adaptions to split-belt treadmill walking in healthy adults

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Declaration

This thesis is the result of the author’s original research. It has been composed by the author and has not been previously submitted for examination, which has led to the award of a degree.

Signed: Adam Booth

Date: 12/08/2015

Acknowledgments

I would first like to thank all the kind people who volunteered their free time to take part in this study, without you all this would not have been possible.

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Contents

Abstract

Disability as a result of stroke is a major issue. Gait is a priority in rehabilitation, where asymmetry is a common and detrimental feature. Conventional treadmills are not able to treat this asymmetry. Recently the use of split belt (SB) treadmills has been suggested as a novel approach, SB allows each leg to travel at independent speeds. This study seeks to investigate the effects of SB gait on healthy young adults as a model for stroke patients. The study carried out motion analysis using the Motek CAREN. Subjects (n=10) were exposed to a total of 15 minutes SB in 5 and 10 minute blocks with a period of normal walking either side. Step length asymmetry and accommodation strategy was analysed. SB resulted in initial negative asymmetry, inferring a shortening of step of the leg on the fast belt (n=6) with some subjects showed a mirrored response (n=4). The asymmetry returns to baseline as subjects accommodate the novel condition. A distinct strategy was characterised, in which step length on fast belt was increased by delaying heel strike. When returned to normal walking conditions the initial asymmetry is reversed with a longer step on the fast leg. This is transient, after a short period normal walking there is a return to baseline asymmetry. Upon subsequent exposure to SB the adaption process is more rapid in adopting heel hang strategy (1st exposure 13 strides, 2nd exposure 3 strides), with significantly reduced asymmetry ($p<0.001$). The results show that young healthy adults can adapt to SB with distinct strategies. The improved adaption as a result of subsequent exposure suggests that there is some level of learned adaption. If applied to stroke patients, long term SB exposure may improve asymmetry and may be encouraged to adopt specific strategies to accommodate an increase in step length.

Introduction

The foundations of rehabilitation science can be pieced together through the early work of many great thinkers and the innate curiosity regarding the human form and movement; from the founding father of anatomy in Galen (131-201 A.D), to the masterful drawings of Da Vinci and Vesalius depicting the human body (14-15th Century). This understanding of the human body paved the way for the development of the study of human movement or biomechanics as it is known today. While a full knowledge of the human form is fascinating, an understanding is demanded to improve function when injury occurs. This can be considered the study of rehabilitation, a crucial field of study that endeavours to improve the recovery of patients. One highly researched aspect, presenting a number of complexities, is rehabilitation of stroke survivors.

Cerebrovascular accidents (CVA) or stroke is one the most serious long term debilitating illness with over 150,000 people suffering from a stroke in the UK every year (Townsend et al., 2012). While incidence rates of stroke has slowly declined in recent years (ISD Scotland, 2015), possibly due to improved lifestyle and the introduction of the smoking ban to Scotland in 2006. Although the extent to which the smoking ban has impacted stroke has not been fully established, a report showed there has been a stepwise reduction in cerebral infarction since its introduction (MacKay et al., 2013). Mortality rates in Scotland associated with stroke sit at around 4,500 per year, as treatment has improved the mortality rate have steadily declined over the years, yet disability remains a probable outcome. Even though incidence declines, it remains the third most common cause of death in Scotland and the most common cause of complex disability in adults (General Register Office for Scotland, 2014).

Stroke commonly results in hemiparesis and a third of those who suffer a stroke are left with a lifelong debilitating illness, resulting in a high burden of care, both socially and financially. Stroke patients account for 7% of all NHS beds and care costs exceed £100million per year, accounting for approximately 5% of the entire Scottish NHS budget (NHS Scotland, 2015; Saka et al., 2009). This figure does not even take into account the wider economic detriment through the inability to work resulting in loss of earnings. The economic loss is far outweighed by the social damage felt by the patient with the inability to fulfil previously held roles. The ultimate goal of rehabilitation is to regain independence and walking is widely considered by patients as one of the most important factors in recovery (Harris et al., 2004). Further to this it is found that regaining independence post stroke, of which ambulation is a critical factor, is closely linked to quality of life and preventing the insidious development of depression (Robinson-Smith et al., 2000).

Stroke often results in complex disability but hemiplegia is a common result, this is the weakness of an entire side of the body, in stroke injury this is contralateral to the site of brain injury. Stroke results in damage to the brain due to an interruption in blood flow. The brain has specific centres of control for example; speech, vision, taste and motor control. The site of damage determines which functional ability is lost or impaired. Motor impairment is the most common functional loss, this affects approximately 80% of stroke survivors (Langhorne et al., 2009). It can be varied in its severity or location but typically impacts motor control of the face, arm and leg of the contralateral side of the body. This is particularly important to gait as walking is a complex co-ordination of muscle activations and feedback on limb positioning and balance. A fitting description of gait parameters in relation to stroke is provided by Onley & Richards (1995). The requirements of gait can be attributed to four aspects: maintaining the balance of the trunk, limbs and arms throughout motion; retaining postural control of limb segments during stance phase; clearing the floor with swinging foot; and ultimately accomplishing all of this with sufficient energy generation and preservation to allow for effective walking. The hemiparetic gait is characterised by slow walking pace, compounded by a decrease in cadence and stride length. Crucially there is also asymmetry, this is manifested in proportions of stance time and step length of each leg. An asymmetric gait is not conducive for normal walking and is considered a clinical tool for assessing stroke rehabilitation (Patterson et al., 2010; Allen et al., 2009). Asymmetry may be important clinically as it has been implicated to be a negative factor in inefficiency, balance control, loss of bone mass in the paretic limb and overuse damage in the non-paretic limb (Jorgensen et al., 2000; Patterson et al., 2008).

Rehabilitation of stroke has received much attention and improvements have been developed with use of functional electrical stimulation and robotically/mechanically assisted motion shows great promise (Mehrholz et al., 2012). NICE guidelines recommend a structured approach that uses a number of techniques. A recent Cochrane review concluded that treadmill training improved stroke rehabilitation over regular physiotherapy intervention (Mehrholz et al., 2014). Although it did not increase the likelihood of regaining the ability to walk it did enhance recovery of waking speed and endurance. This improvement although small is suggested to offer significant long term clinical improvements in quality of life. While conventional treadmill training has been shown to offer promising results in rehabilitation it is not possible to address the difficulties asymmetry presents to the stroke patient.

Early investigations into motor control pathways showed that decerebate cats were able to adapt to a split belt treadmill (Graham-Brown, 1911). This featured two independent belts that were capable of travelling at differing speeds, cats were able to respond to the changed conditions of the split belt, adopting altered walking patterns. The implications of this research highlighted the possibility that adaptive motor control was not the sole responsibility of the higher areas of the brain such as the primary motor cortex. As the human nervous system is significantly more complex, there were questions as to the suitability of animal studies as models for humans in motor control pathways. Preliminary research by Dietz (1994) and Prokov et al. (1995) investigated the effect of split belt conditions on humans. It was found that healthy individuals were able to adapt to the novel condition. It was commented by participants that the sensation of split belt condition persisted after the session ended, basic EMG was also collected that showed that some level of gait adaption of motor co-ordination. The inferences from this work eventually led to a question over its possible use as a rehabilitation tool. Consequently, more recent studies have investigated the beneficial effects of the use of SB in rehabilitation (Riesman et al., 2005; Morton et al., 2006; Reisman et al., 2007; Huynh et al., 2013). It was found that on SBT, inter-limb spatiotemporal parameters of gait such as step length and double support time can be adapted; these effects are reported to persist in the opposing limb transiently when returned to normal treadmill conditions as the body overcompensates for the perceived change (Reisman et al., 2005). As animal models have shown, this is likely not a function totally reliant on the primary motor cortex. This is important in stroke recovery as this area is often the site of trauma. Indeed further studies by Reisman et al. (2007) found stroke patients with cerebral impairment could adapt in the similar extent as healthy subjects. Training under these conditions then resulted in improved symmetry when returned to normal treadmill conditions (Reisman et al., 2009). Repeated long-term training has been shown to result in prolonged retention of improvements to gait symmetry (Reisman et al., 2013).

The use of split belt treadmills has been investigated to a limited degree and much of the research has been carried out by one laboratory at the John Hopkins Medical School, Baltimore. There is a lack of clear data reporting the spatiotemporal modifications of gait that occur during the adaption to split belt walking. The aim of this investigation is to add to the current knowledge regarding the effect of split belt walking and the implication this may have on motor control and rehabilitation of stroke.

This study will seek to examine the changes in spatial temporal gait parameters that are observed when healthy subjects, novel to split belt, adapt to the SB condition. The study will also seek to investigate the reported after-effects that may be present when subjects are returned to a normal treadmill walking condition after prolonged split belt walking. It will investigate a dose-response action through changes in exposure time to split belt walking. Further to this, as there are implications related to motor learning, the study will explore the learning effect by exposing each participant to a second spell of split belt walking and identify any evidence for a learned response through improved adaption.

The research will be the first of its kind to gain highly accurate and detailed biomechanical data through its use of gold standard motion capture system in the form of Vicon cameras (Vicon, Oxford, UK). It will be further unique in its use of a Computer Assisted Rehabilitation Environment (CAREN) (MotekForceLink, Amsterdam Netherlands). The system incorporates a split belt treadmill with harness rig on a platform capable of movement in 6 DoF, an immersive environment can be created by use of 180^{o} screen projection and surround sound. As this is a pilot study, using healthy adults as model for stroke patients, the use of split belt in the CAREN environment as a rehabilitation tool will be considered.

It is hypothesised that introduction to split belt treadmill walking will cause initial difficulty for subjects, creating an asymmetry in step length, as a result of a reduction in step length of the limb on the fast belt. An adaption process will occur in which step length symmetry will return towards to a baseline as subjects accommodate the novel walking condition. When the subjects are returned to normal walking after a period of exposure to split belt it is predicted that asymmetry will then persist for an indeterminate amount of time on the opposite side. This will be caused by an increase in step length of the limb on the fast belt as the body overcompensates.
It is also hypothesised that upon repeated exposure to split belt walking, a level of motor learning will occur and accommodation to split belt conditions will be more rapid in returning to baseline step length symmetry.

Literature Review

Brain Anatomy & Pathology

The brain can be considered the most vital organ in the body and highly sensitive to metabolic change. Failure of blood supply is therefore a catastrophic event. This is described as a cerebrovascular accident, or more commonly termed a stroke. Loss of blood to the brain can result from either occlusion of a vessel or haemorrhage of a vessel, referred to as ischemic and haemorrhagic strokes respectively. Both of these cause an interruption of blood supply to the brain tissue or build-up of hydrostatic pressure, resulting in death of neural tissue with serious consequences.

Ischemic strokes are the more common type of stroke, they account for approximately 85% of strokes across the UK (Luengo-Fernandez et al., 2013). These are caused by occlusion of the arteries in the brain, usually as a result of thrombosis. The brain has a complex network of blood supply that provides sustenance for different regions of the brain. Therefore the location of the occlusion is key to defining what portions of the brain suffer damage. The middle cerebral artery (MCA) is a common site of ischemic stroke and has been used in many experimental animal models (Bederson et al., 1986; Longa et al., 1989). The MCA is important as it carries blood supply to critical parts of the brain such as speech, comprehension and crucially in the context of this thesis, parts of the motor cortex (Gray's Anatomy, 2010).

Transient ishemic attatcks (TIA) can be considered mini strokes and are a strong precursor for a stroke. TIA present as stroke like symptoms but only present for under 24hrs. Each year 46,000 people have an incidence of TIA (Scarborough et al., 2009). Rothwell et al. (2010) noted that 11% of patients presenting with TIA went on to develop a full cerebrovascular accident within one week, in the same report they also noted that variable blood pressure was a more predictive risk factor for stroke than high BP alone. This is theorised to be the due to the damage caused by of stretching and slackening of the often stiffened arteries of at risk groups.

Haemorrhagic strokes are not as common as ischemic stroke but account for a considerably higher mortality rate of over 40% (Andersen et al., 2009). Haemorrhagic strokes can be categorised by their area of action.

An extradural haemorrhage is a pooling of blood between the skull and dura resulting in increased hydrostatic pressure on the brain. It is generally caused by trauma to the head resulting in a rupture of vessels (Gray's Anatomy, 2010). In this case the patient may be asymptomatic for a number of hours before deteriorating rapidly, hence why all head injuries should be treated with great caution even if minimal symptoms present.

Subdural haemorrhage is the result of bleeding from the cerebral veins, it may be deemed acute or chronic. Acute injuries occur in high accelerations/decelerations forces such as those experienced in a car crash, inertial forces cause movement of the brain resulting in rupture of the connecting vessels. Chronic subdural haemorrhage is more commonly associated with elderly patients suffering trips or falls. In the elderly, the intra cranial vessels are shrunken and stiffer where small knocks can cause slow bleeding in the brain. Symptoms may not present for a number of days or weeks but often results in episodes of confusion, dizziness or disorientation (Gray's Anatomy, 2010).

Stroke Epidemiology

In the UK each year 152,000 people suffer a stroke with approximately 15,000 of those occurring in Scotland (Townsend et al., 2012). It is one of the leading causes of death and disability in the modern world. It is a condition often associated with old age and therefore as women live longer they account for the majority of stroke incidents overall. For aged matched individuals however, men are more predisposed at a younger age (Townsend et al., 2012). The links with ageing and stoke are well defined, it has been shown that the risk of stroke doubles every decade over the age of 55 (Brown et al., 1996).

Both incidence and mortality of stroke worldwide has decreased since 1990 by 12% and 37% respectively (Feigin et al., 2013). It was also noted in the study by Feigin et al. (2013) that incidence of stroke in a younger demographic (20-64) was worryingly high, increasing by 25% worldwide in recent years. This was particularly prevalent in low and middle income countries although reasons for this were not suggested. A similar trend confirms this within the UK showing that more socially deprived areas show a doubled incidence of stroke (Public Health England, 2015). With increasing life expectancy, younger incidence and improved treatment this has led to a sharp increase in "disability adjusted life years (DALYs) lost", i.e. a prolonged burden of care.

Ethnicity has been shown to play a role in stroke prevalence. Individuals of South Asian and Afro-Cariban decent display higher incidence of stroke than individuals of white European origin (Wang et al., 2013). This is considered to be due to the predisposition to diabetes and high blood pressure. High blood pressure is becoming particularly prevalent in western countries with growing sedentary lifestyles and poor diet choices. The lifestyle and diet is also contributing to a steady increase in the diagnosis of type 2 diabetes. Both of these conditions increase risk of stroke (Kothari et al., 2002). High blood pressure places increased pressure on the cardiovascular system and the prolonged exposure to high glucose levels as a result of diabetes contribute to the build-up of plaque on the vessel walls (Dyken, 1991). There are several other risk factors associated with lifestyle that have been identified with stroke prevalence that are generally considered to contribute to increasing the cholesterol levels such as a diet that is high in saturated fat, smoking, lack of exercise, high alcohol intake, liver and/or kidney disease. A wide number of reports suggest that many of these strokes can be prevented by improved lifestyle and dietary choices (WHO, 2009; Strazzullo P, et al. 2010; Naci H et al. 2013).

Stroke is the 4th leading single cause of mortality in the UK (UKGov, 2014). As stated, mortality as a result of stroke is decreasing, this has led to an increase in number of stroke survivors. Currently 2.16% of the Scottish population are stroke survivor (ISD Scotland, 2014). Approximately half of all stroke survivors are left with long term disability, meaning it is the single largest cause of complex disability (Adamson, 2004; SSNAP Audit, 2014). Table 1 shows some of the most common difficulties that stroke survivors face in the months and years following stroke.

Difficulty	% of stroke survivors affected
Upper limb weakness	77 %
Lower limb weakness	72%
Visual problems	60%
Facial weakness	54%
Slurred speech	50%
Swallowing	45%
Sensory loss	33%
Depression	33%

Table 1. Common disabilities that impact on stroke survivors. *Figure adapted from Stroke Association, (2015)*

It is estimated that the socioeconomic impact of stroke in the UK is up to £9 billion per year (Saka et al., 2009). While this number may be important in governing decisions, on a human level this does not compare to the daily impact of stroke in the community. The burden of care often falls on friends and family, resulting in high stress and anxiety for all parties that may lead to further mental health problems.

Rehabilitation

To reduce the impact of stroke, rehabilitation is a central part of recovery. There are specific guidelines of best practise, based on extensive research evidence, laid out by the Scottish Intercollegiate Guidelines Network (SIGN) (2010) and National Institute for Health and Care Excellence (NICE) (2013).

These provide recommendations for specific pathways of stroke rehabilitation. The scope of disability is wide ranging, therefore, each patient first needs to be screened and assessed paying particular attention to vision, hearing, muscle tone, strength, sensation and balance. Planning of rehabilitation and goal setting is encouraged to be tailored to each individual with both long term and short term goals set. The goals must be challenging to the patient but ultimately achievable, involvement of family and friends in these goal settings has been shown to increase effectiveness (Langhorne, 2011). The main aim of rehabilitation is to regain independence; often regaining independent walking is one of the most highly prioritized selections (Bohannon et al.,1991). For many people, the re-integration into community life marks the end point of their rehabilitation. The precise definition of community ambulation is somewhat ambivalent but certain definitions are widely used. Gait speed, timed up and go, and walking endurance are all common and useful measurements. Lord et al. (2005) argue that these measurements do not encompass the requirements of community ambulation, highlighting the importance of attention on outdoor walking ability. This may be important in the present study as the use of immersive virtual reality and visual flow may allow for the development of this aspect in future rehabilitation strategies.

Fitness training, repetitive task training and strength training are all key processes in the early stages of motor rehabilitation that are carried out with the physiotherapist. A wide range of motor learning and neurophysiological techniques are also purported, with the Bobath treatment (Bobath, 1990) the most commonly used in Europe, it works on the principal that muscle weakness is caused by antagonist spasticity. Physiotherapists physically mobilise the spastic muscles attempting to reduce the spastic tone. There is a wide variety of these techniques reported (Belda-Lois et al, 2009), but there has been no evidence found to support the use of any one of these techniques alone and do not offer any improvement of outcome compared to typical rehabilitation such as traditional exercises and functional training (Van Peppen et al., 2004). SIGN guidelines (2010) place the highest emphasis on gait-oriented physical fitness training and electromechanical devices to improve functional ambulation for patients assessed as medically stable and functionally able to participate. Indeed the combination of different rehabilitation strategies would appear to be more effective than over-ground gait training alone. Recent advancements in robotics and mechanically assisted stepping devices have been reported to show great promise in rehabilitation of gait re-training (Merholz et al., 2012). They may be in the form of robotic exoskeletons or mechanical gait simulators, both support the body to simulate a normal gait, allowing for high repetition of movement that may stimulate gait retraining. Functional electrical stimulation (FES) is another technique that may be of future interest in rehabilitation, shown to illicit clinically significant improvements in muscle strength and gait (Glanz et al., 1996). FES is a process of electrically stimulating muscles at precise times to create movement, simulating the effects of an action potential to a motor unit. It may be of particular benefit with conditions such as drop foot, where the dorsiflexors may be recruited by electrical excitement at specific timing to prevent drop foot occurring.

Ankle foot orthoses may be used to improve walking speed efficiency, gait pattern or weight bearing during stance (SIGN, 2010). NICE guidelines also recommend electromechanical gait stimulation and walking therapy. It is the latter that will be the focus of this investigation.

Gait retraining improves with practice (Langholme, 2009). Treadmill training allows for a large number of steps to be taken in a controlled, safe environment. It is widely used in rehabilitation due to its ease of access. Many studies have investigated the beneficial effects of treadmill training with and without body weight support (Merholz et al., 2012). Body weight support is an innovation developed from animal models with spinal injury (Barbeau & Rossignol, 1987; Lovely et al., 1986). It is suggested that unloading of the lower limbs, most commonly by means of harness, allows for an environment that discourages the recruitment of compensatory movement strategies that may worsen asymmetry in gait (Sullivan et al., 2002). Early studies showed great promise in the technique (Visintin et al., 1998). Since then the large number of studies have elicited some variation in outcome. A wide ranging Cochrane review, assessing 44 clinical trials, found that treadmill training both with and without body weight support did not did not increase the chances of an individual returning to independent walking compared with other physiotherapy interventions in stroke patients (Merholz et al., 2012). For those that were ambulatory, however, there was found to be a clinically significant improvement in both walking speed and endurance, with a pooled mean improvement of 0.07m/s (p=0.02) and 26.35m (p=0.03) respectively. It was found that those who were most ambulatory benefited most from the measure, the extent of the improvement varied between each individual as expected. In a similar meta-analysis study van de Port et al. (2007) found that gait related treadmill interventions improved walking speed and endurance to an even greater extent. When considering treadmill training without body weight support a further systematic review by Polese et al. (2013) suggested even more encouraging results with a mean improvement of 0.14 m/s walking speed and 40 m greater distance endurance and that these effects persisted beyond the intervention period. It was even concluded that these results were a "conservative" estimation of the benefits of treadmill training. It is considered that walking speeds of 0.4-0.8m/s allow for good community ambulation (Duncan et al., 1998). Therefore what can be considered a small improvement in gait speed is a proportionally high improvement.

Given the developments in rehabilitation strategies over recent decades a large number of people, up to 60%, are still unable to walk confidently in the community a number of month post stroke (Langhome et al., 2009). It is therefore imperative that improved methods of stroke rehabilitation must be investigated. In order to develop methods of rehabilitation the neuromuscular mechanisms underpinning motor control must be understood.

Neuromuscular Control

The normal gait is an intuitive movement for most, but to consider it in detail it is a highly complex system that is learned and perfected over a number of years. Gait is a mixture of voluntary and automatic actions that work together to form an individual's gait. In order to achieve this, the control system, i.e. the brain, must have an innate knowledge of precisely where the limbs are at any given time and the forces acting on them. This sensory input may be provided by the visual field, however, the innate position of the limb may be known without any other sensory input. This is known as proprioception and can be considered a sixth sense. It is in the form of sensory receptors, namely; muscle spindles, Golgi tendon bodies and mechanoreceptors receptors in joints, subcutaneous tissue and skin (Rosenbaum, 2010).

Muscles exert a force as a result of electrical stimulation from nerves. Muscle fibres are innervated by a single motor neurone, but each neurone may innervate numerous muscles fibres, this is known as the motor unit. The number of muscle fibres innervated in a motor unit varies. In muscles requiring fine control of movement such as the hand and eye, fewer fibres are inverted by a single a motor unit compared to those in the lower leg, where a single motor unit may involve as many as 10x the muscle fibres (Buchthal & Schmalbruch, 1980). The type and number of motor units activated in a contraction of a muscle is dependant of the force required and the firing rate of the incoming signals. This maintains efficiency as large motor units will not be required to be recruited at lower forces (Rosenbaum, 2010). The model of recruitment of motor units is a topic of some debate and speculation with different theories contested (Henneman, 1957; De Luca & Contessa, 2012; De Luca, 2015)

Motor control arising from the brain can be considered as communications between three major regional areas; the cerebral cortex, cerebellum and the basal ganglia (Rosenbaum, 2010). The cerebral cortex contains the primary motor cortex. The function of this area was displayed in early work by Fritsch and Hitzig (1870), with electrical stimulation to specific regions resulting in muscle twitches of localised parts of the body. A topographic map of the motor cortex can then be made. These regions stimulate movement of the muscles, thus the higher control centre. There are additional motor areas, the secondary motor cortex and supplementary motor cortex. These have more complex actions including orientation, high-level planning and production of complex movement sequences (Rosenbaum, 2010).

The basal ganglia are a group of interconnected regions of the forebrain, namely the caudate, putamen and globus pallidus (cumulatively forming the lenticular nucleus). The role of the basal ganglia is complex as a result of their highly interconnected nature, however, they display two distinct pathways; direct and indirect (Keinirim, 2007). While the function of the basal ganglia is complex, an insight into the actions it provides can be ascertained by investigation of diseases that affect it. Huntingtons disease involves death of dendrites within the basal ganglia, it results in choreiform movements (involuntary, continuous movement of the body) particularly at the extremities and face. Parkinsons disease results from the loss of domaniergic neurones in the signalling pathway of the basal ganglia. This condition has a somewhat opposite effect with bradykinesia or akinesia (slowness or absence of movement) usually accompanied with a resting muscle tremor (Rosenbaum, 2010). These degenerations of the basal ganglia demonstrate it's important in the motor control system.

The cerebellum is a crucial part of the brain, responsible for co-ordination and plays a role in learning complex motor tasks (Glickstein, 2007). Again much of our knowledge of brain function arises from brain injury. Lesions of the area can result in ataxia, a co-ordination disorder and difficulties in timing of movements (Rosenbaum, 2010). In this respect it could therefore be considered as a filter, compensating changes in input from the higher brain and feedback from limb, fine-tuning activation to give more precise, steady movement. The cerebellum is highly connected to all areas of the brain and therefore a full understanding of its role is difficult to pinpoint. Early studies on decerebate cats, i.e. lesion of the cerebrum, showed images of walking on treadmill in 3 different gait styles in response to speed changes (Graham-Brown, 1939). This suggests that certain learned motor skills are not reliant on higher motor input. In a detailed review Whelan (1996) describes the role of the cerebellum in detail but warns that even simple motor tasks of walking round a corner require significantly more control than displayed in the decerebate cat, concluding that the "breathtakingly complex nature of the brain" cannot be broken down into simple unit blocks. While animal studies may provide an insight the human brain is more complex and differently structured, not to mention the immense complexity of the phenomenon surrounding brain plasticity and learning.

Motor Learning

Due to the immense complexities of the brain, the topic of motor learning is one that is continuously researched and updated with an array of different theories regarding its nature (Latash & Lestienne, 2006). "Motor learning does not need to be rigidly defined in order to be effectively studied", this statement was made in regard to motor learning and stroke rehabilitation (Krakauer, 2006). To some extent this is true but basic the processes of motor learning needs to be understood in order to maximise re-learning of motor skills affected during brain injury. The sole performance of an action does not imply learning, this must consider the development of an action over time, observing characteristics such as improvement, consistency, stability, persistence and adaptability (Magill, 2007).

Early understanding of motor learning worked on the principal that practise makes perfect (Latash & Lestienne, 2006). This may in basic principle hold true but it does the complexity of the topic injustice, consider just one of the proposed models of learning; the three-stage model set out by Fitts and Posner (1967). During the first stage, the cognitive stage, the subject new to the motor skill attempts to work out the objective of the action, move what, where and when. They require a high level of cognitive thought to co-ordinate individual movements to carry out the action. The cognitive stage is discernible by it high error and variability. The second stage, the associative stage, occurs as the subject identifies the correct co-ordination of movements and links this with timing relating to environmental cues. There is a reduction in error and variability as the movement is refined. Eventually the action may progress to the autonomous stage. At this point the movement is habitual and can be carried out with minimal conscious thought, there is high accuracy and consistency as the movement is highly refined. Walking can be considered at this autonomous level among healthy subjects. There is not a definitive length of time, or number of practise sessions that defines these stages, they may be influenced by personal traits and can be influenced by instruction and augmented feedback (Magill, 2007).

Gait

Walking could be argued to be the most important developmental skill of the human species; it is thought to have been crucial to the evolutionary success of our early ancestors (Bramble & Lieberman, 2004). The first evidence of the bipedal hominid gait appears around 3.6 million years. Tracks left in volcanic cement, found in Tanzania, of what are thought to be a parent and child walking side by side, show footprints that have very similar anatomy to that of man today; with a straight hallux and rounded heel. A great number of people have sought to characterise the walking pattern and a plethora of books and research papers give each individuals viewpoint on what could be considered a basic movement. This is poetic in that gait itself is a diverse and unique movement; an individual can be identified from a distance by their walking style.

There are mechanical theories that help to underlie the properties of the gait cycle. There are two main theories; the inverted pendulum theory (Cavagna & Margaria, 1966) and the determinants of gait theory (Saunders, Inman, & Eberhart, 1953). The pendulum theory states that the leg acts as an inverted pendulum with a straight leg planted on the ground the centre of mass (COM) travels over in the arc of a pendulum, a highly efficient mechanical principle.

The determinants of gait theory revolves around the centralisation of the COM, where knee, pelvic and ankle movements are used to reduce deviation of the COM. This theory has been widely accepted over the years and is described in many textbooks (Perry, 2010; Levine, 2012), but under experimental investigation it has been discovered to be flawed (Kuo, 2007). Similarly but to a lesser extent the inverted pendulum theory is also flawed, instead, dynamic walking theory has been described as an alteration of the inverse pendulum theory (Kuo, 2007). The single leg phase acts as an inverted pendulum but the model includes the work required in the step to step transition where the leading and trailing legs must perform negative and positive work respectively. The model seems the most appropriate description of the gait cycle.

The walking gait can be broken down into two basic periods; stance and swing period (Fig. 1). Stance period is described as the period of time that the foot is in contact with the ground, while swing period can be considered the time in which the foot is in the air. Stance period can further subdivided into phases: initial double limb stance, single support limb and terminal double support stance. The proportions of these timings are related to walking style and speed. Speed has an inverse relationship with phases, an increase in speed decreases the time at both phases, however faster walking proportionally increases single support time and reduces the double support time. It is no longer considered walking when both feet are considered in swing phase (i.e. no portion of the body is in contact with the ground) this is defined as running (Levine et al., 2012).

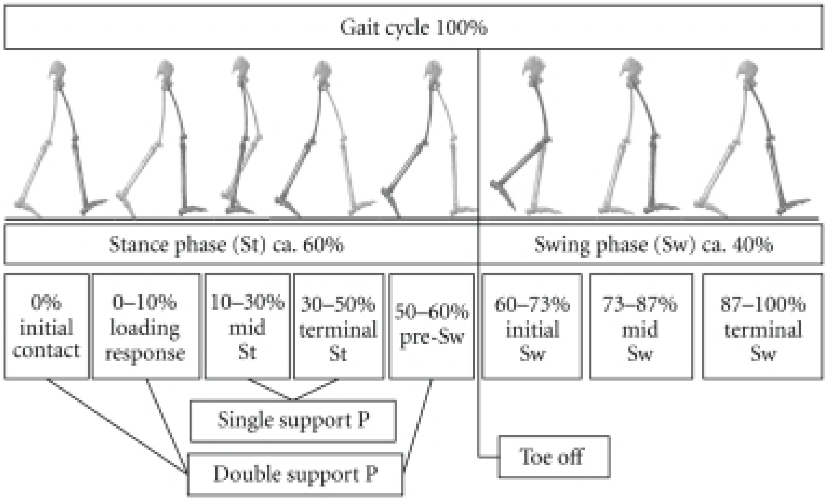


Figure 1. The gait cycle showing different phases and proportion of cycle in which they occur. *Figure adapted from Levine et al., 2012*

Again each phase can be broken down into further phases and events. We will consider the temporal progression of one leg from initial stance phase to the repeating cycle, giving stride length. The first phase can be considered as initial contact, generally observed as a heel strike. At this stage the foot impacts the floor, leading to foot drop (plantar flexion) and initiates the transfer of the body weight. The phase progresses to the loading response phase, here the force is loaded onto the leg with the knee flexing to provide some shock absorbing and soft tissue spring loading. Mid-stance occurs as the opposite leg is raised off the ground, at this stage the entire body weight is on the single leg. As the centre of mass is not directly over the leg this is a period of greatest instability and muscles in the trunk and leg are required to maintain balance. Progressing to terminal stance the stance leg extends at both the hip and knee, the ankle becomes plantarflexed and momentum along with soft tissue elastic tension propel the centre of mass forward.

The swing period can also be segregated into phases. Pre-swing directly follows terminal stance as the knee goes into flexion and ankle further plantarflexes. The leg then goes into initial swing phase; the hip and knee flex and ankle dorsiflexes to allow the foot to clear the ground with progression of swing. During mid swing the foot is parallel with the ground and gravity brings the tibia to the vertical. Finally terminal swing occurs with extension of the knee with ankle remaining neutral, readying for heel strike and progression of gait.

A definition of terms is useful at this point. Gait can be defined by variables, these can be categorised as spatial-temporal parameters. There are countless variables that may be measured, however, in the interest of brevity only the following will be described and discussed:

Walking speed is simply the overall speed at which the subject walks, this is most commonly quoted in m/s with a general average for comfortable self-selected over ground walking speed reported between 1.1 – 1.4m/s (Oberg, 1993). This is a widely variable parameter and individuals can vary with age, height and personal preference.

Cadence is defined as the number of steps per minute. Again this is variable between individuals but reference data suggests an average between 110-120 steps/min (Perry, 2010)

Stance time is the total length of time spent during stance phase of gait cycle on one leg. Single limb support time is the length of time during the stance phase in which it is the only supporting leg is in contact with the ground. Double support time is the length of time spent with both feet in contact with the ground during one gait cycle.

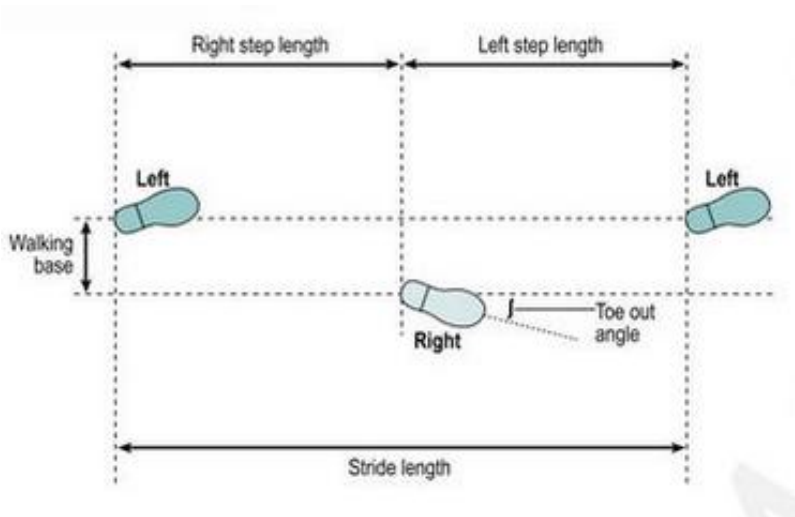


Figure 2. Step and stride length in gait cycle. *Figure adapted from Levine et al., (2010)*

Step length can be defined as the distance from one foot to the other during double support phase. Stride length can be considered as the distance between placements of the consecutive leg (Fig. 2). The definition of this becomes somewhat more complex when investigation gait on a treadmill as the foot is not on a set position on the ground.

Gait is often described by its kinematics, i.e. the angular positioning of the body segments around a joint; Fig. 3 shows this in detail. The displacement of these body segments are given different terms depending on their plane of action. Movement in the sagittal plane it is referred to as flexion and extension. Coronal plane movements are known as abduction and adduction. Finally in the transverse plane internal and external rotation takes place.

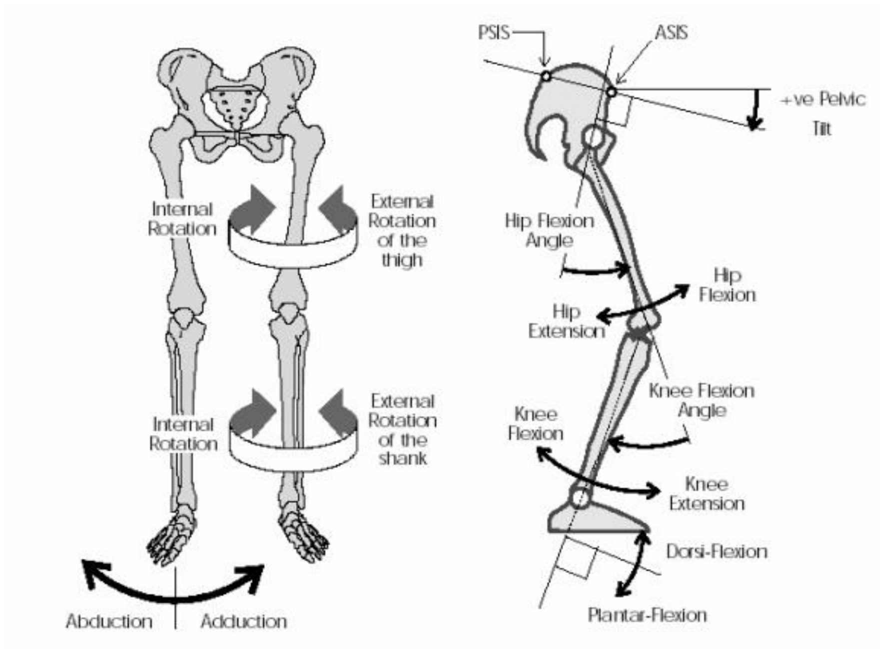


Figure 3 Coronal (left) and transverse (right) diagrammatic views showing the joint axis and direction of movements of flexion/extension, ab/adduction and internal/external rotation. *Figure adapted from Perry et al., (2010)*

Much of the work towards the understanding of human gait has been conducted with the use of treadmills. Treadmills allow for a large number of gait cycles to be recorded for assessment in a controlled environment giving greater reliability. In rehabilitation this is also very useful as it allows for a high volume of practice repetition which is a key feature of motor learning. The use of treadmills however presents some problems as it illicit small alterations in gait.

Over Ground vs. Treadmill Gait

There have been questions over the comparability of treadmill walking vs. over ground walking due to the subtle difference in motion, summarised in Fig. 4. A common point noticed is that when walking on a treadmill a subject will choose a slightly slower speed than would be selected when walking over ground (Riley et al., 2007). Other studies have sought to investigate the kinematic differences further showing changes in maximum hip flexion and knee extension which were more pronounced in treadmill walking (Alton et al., 1998; Parvataneni et al., 2009; Watt et al., 2009). The studies generally conclude that the changes are small and tolerable with the benefits of the use of treadmills to carry out a large number of repeated steps in a given space outweighing the slight changes in gait parameters. Indeed, Lee et al (2007) concluded: "While differences were observed in muscle activation patterns, joint moments and joint powers between the two walking modalities, the overall patterns in these behaviours were quite similar. From a therapeutic perspective, this suggests that training individuals with neurological injuries on a treadmill appears to be justified."

A

	TM	OG
Cadence (steps/min) (S.D.)	113.70 (8.11)	114.13 (8.30)
Walking speed (m/s) (S.D.)	1.41 (0.16)	1.48 (0.18)
Stride time (s) (S.D.)	1.06 (0.08)	1.06 (0.08)
Opposite foot off (% cycle) (S.D.)	10.65 (1.67)	11.02 (1.47)
Opposite foot contact (% cycle) (S.D.)	50.02 (0.82)	50.10 (0.99)
Foot off (% cycle) (S.D.)	60.59 (1.58)	61.17 (1.57)
Single support (% cycle) (S.D.)	41.67 (2.15)	41.22 (2.80)
Double support (% cycle) (S.D.)	22.66 (4.53)	23.04 (4.58)
Stride length (m) (S.D.)	1.48 (0.12)	1.55 (0.13)

B

	Mean difference	S.D.	*t*-Test
Hip flexion	**0.64**	**1.31**	**0.02***
Hip extension	**−1.50**	**1.22**	**0.00***
Hip abd	0.02	0.98	0.92
Hip add	**−1.41**	**0.96**	**0.00***
Hip ext rot	0.38	1.63	0.25
Hip int rot	**−2.28**	**1.95**	**0.00***
Knee flexion	0.68	1.74	0.06
Knee extension	**−0.58**	**0.92**	**0.00***
Ankle plantarflexion	0.83	5.40	0.44
Ankle dorsiflexion	−1.69	5.92	0.16
Pelvic tilt anterior	−0.46	1.31	0.08
Pelvic tilt posterior	**−0.92**	**1.38**	**0.00***
Pelvic obliquity max	**0.90**	**0.70**	**0.00***
Pelvic obliquity min	**−0.77**	**0.73**	**0.00***
Pelvic rotation max	**1.04**	**1.07**	**0.00***
Pelvic rotation min	**−1.47**	**1.44**	**0.00***

Figure 4A/B A) Differences in treadmill vs over ground walking. Note slower walking speed, reduced cadence and stride length of TM. B) Kinematic differences of treadmill vs. over ground walking.* **BOLD** indicate significant difference. TM- treadmill. OG- Overground.

Figure adapted from Riley et al. (2007)

The Hemiparetic Gait

One of the most common symptoms post stroke is hemiparesis, i.e. weakness of muscles on the body side contralateral to the lesion. This causes significant problems with motor control and gait. While general characteristics can be observed, due to the specific nature of the location and extent of each brain injury, the outcomes cannot be easily grouped. A fitting description of gait parameters in relation to stroke is provided by Onley & Richards (1995). The requirements of gait can be attributed to four aspects: maintaining the balance of the trunk, limbs and arms throughout motion; retaining postural control of limb segments during stance phase; clearing the floor with swinging foot and ultimately accomplishing all of this with sufficient energy generation and preservation to allow for effective walking. The hemiparetic gait is characterised by slow walking pace, mirrored with a decrease in cadence and stride length. Crucially in regards to this body of work there is also asymmetry, this is manifested in proportions of stance time and step length of each leg.

Walking speed is often greatly reduced with both stride length and cadence suffering. Differences in the proportion of stance and swing phase have been reported. The overall time spent in stance phase is increased. There is a favouring of proportionally prolonged stance on unaffected side over the affected side. There is also a greater proportion of the gait cycle spent in double support. It is noted, however, that when compared to able bodied subjects walking at similar speeds, the hemiparetic subjects' show decreased double support time (Richards, 1996).

An asymmetric gait is not conducive for normal walking and is considered a clinical tool for assessing stroke rehabilitation (Patterson et al., 2010). Asymmetry may be important clinically as it has been implicated to be a negative factor in inefficiency, balance control, loss of bone mass in the paretic limb and overuse damage in the non-paretic limb (Jorgensen et al., 2000; Patterson et al., 2008; Allen et al.,2011). The importance of the emphasis on asymmetry has been questioned by Olney & Richards (1996) as each side has unequal power capabilities they cannot be expected to perform to the same level. This statement may have its merits but the normal walking gait revolves around symmetry and if the end goal is to return to a normal walking gait this should be a major focus of rehabilitation. Conventional treadmills cannot be used to treat asymmetry as each leg is under the same conditions and therefore perpetuating the asymmetry. A novel approach must therefore be considered to improve asymmetry in gait, one possible route for this lies in the use of split belt treadmills.

Split-belt

Split belt (SB) treadmills feature two separate parallel belts that are capable of travelling at independent speeds. Each leg can therefore be driven at different speeds, at a range of ratios from 1:1 up to 1:4 (0.5m/s:2m/s), speeds of 1:2 are most commonly used and show similar response to higher speed ratios while being more acceptable to subjects, with less risk of trips or falls. The interest in motor control and the effects of SB treadmill training has stemmed from studies investigating de-cerebrate cats (Graham-Brown, 1939; Yanagihara et al., 1993). They were found to adapt to SB treadmill speeds showing the cerebrum was not vital in the simple action of reacting to underground conditions. This effect was further investigated in human subjects walking on a circular treadmill. After a period of circular treadmill walking subjects were blindfolded and asked to walk over ground in a straight line. All subjects travelled in a curved trajectory while under the impression they were travelling straight (Gordon et al., 1995).

More recently there has been further investigation into the effects of split belt treadmill walking on neuromuscular control and co-ordination (Deitz et al., 1994; Prokop et al., 1995; Riesman et al., 2005; Morton et al., 2006; Reisman et al., 2007; Huynh et al., 2013). Early studies by Deitz et al. (1994) investigated healthy subjects using split belt at a range of speeds, it was found that healthy subjects were able to adapt within 15-20 stride cycles although the exact definition of the adaption state is not defined other than a "relative shortening of the duration of the support and lengthening of the swing phase of the "fast" leg and, vice versa, in support and swing duration on the "slow" leg". Figure 5 shows the general theory of split belt adaption, with step symmetry being accommodated by spatial temporal adaptions.

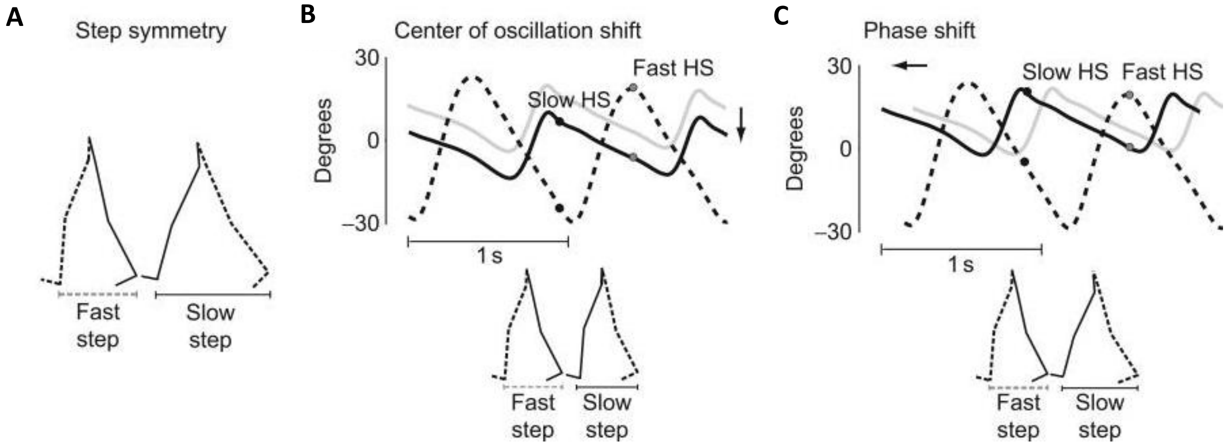


Figure 5. Dashed line represent fast leg, solid line slow leg. (A) An example of kinematic data of two consecutive steps in the early adaption phase is shown. Note the difference in step length, resulting in asymmetry. (B)/(C) show the proposed method of reducing asymmetry. (B) Limb angle trajectories plotted as a function of time in split belt adaptation—two time periods are displayed. Grey line represents the movement in the slow limb in early adaptation. Positive limb angles are when the limb is in flexion. Step lengths can be equalized by changing the position of the foot at landing. This spatial adaption is described as a shift in the "centre of oscillation". (C) Step lengths can also be equalized by changing the timing of foot landing, as shown by the change in phasing of the slow limb from the grey trajectory (early adaptation) to the black trajectory (late adaption). This is described as "phase shift" as subjects equalize step lengths by changing the timing of foot landings with respect to each other. *Figure adapted from Malone and Bastian (2014).*

There is reported to be two distinct changes that occur when walking on a SB treadmill (Reisman et al., 2010). The first change relates to the immediate accommodation of the subject to the SB condition resulting in the slower leg spending a prolonged time in stance phase and the faster leg spending a reduced time in stance phase. This is described as a reactive adjustment and re-adapts just as quickly when belts are returned to the same or "tied" speed. The second change is described as "adaptive, feed forward in nature" (Reisman et al., 2010). This adaption occurs more slowly, aiming to reduce the initial asymmetry between the limbs. Parameters associated with this include step length, double support time and inter-limb phasing. These then persist for a time in the opposite limb when the belts are returned to tied speeds; a possible overcompensation for the learned adaption (Reisman et al., 2005). The studies provide an insight to the complex nature of motor control that deserves to be studied in greater detail, not least because of the possible benefits to rehabilitation.

As previously described the adaptive ability of the brain may be retained in stroke subjects. Indeed preliminary tests of SB treadmill in stroke patients found that it was a feasible with stroke patients showing the ability to adapt (Reisman et al., 2007), the same was also found in subjects following hemispherectomy (Choi et al., 2009). For healthy subjects, the walking patterns adopted are asymmetric in step length (i.e. the normalized difference in step sizes of the two legs) and double support time. When belts are returned to the tied speed this adaption is reversed. If stroke patients are displaying an asymmetric gait then the aftereffects of this exposure could improve symmetry. The asymmetric pattern of the stroke patient is exaggerated during the split belt walking period, therefore resulting in improved symmetry for the individual when returned to normal conditions and the neuromuscular control system attempts to correct the exaggerated asymmetry. The correct side to place on the fast or slow belt is not intuitive as patients with hemiparesis can show asymmetry in either direction. To improve symmetry in walking this must be considered. Essentially the leg with the proportionally shorter step length needs to be trained on the "fast" belt. Consequently if this is incorrectly assigned the step length asymmetry will be exacerbated by SB training. It has not been considered by any researcher the beneficial effects of forcing increased use of the paretic leg, this may lead to an increase in strength of the limb, however, as suggested may result in a more asymmetric gait that may present a long-term problem.

In the early work by Prokop et al. (1994) it was found that on repeated exposure to split belt walking after short break (1 minute) the "adaption" period was greatly reduced from 15-20 stride cycles to 1-3. Again, there lacks clarity in the precise definition of the steady state of adaption, however, this highlights that the motor action is being learned or at least partially stored, with the suggestion that further training would lead to better results. Figure 6 shows the adaption curve in response to the 1st and 2nd exposure to split belt condition.

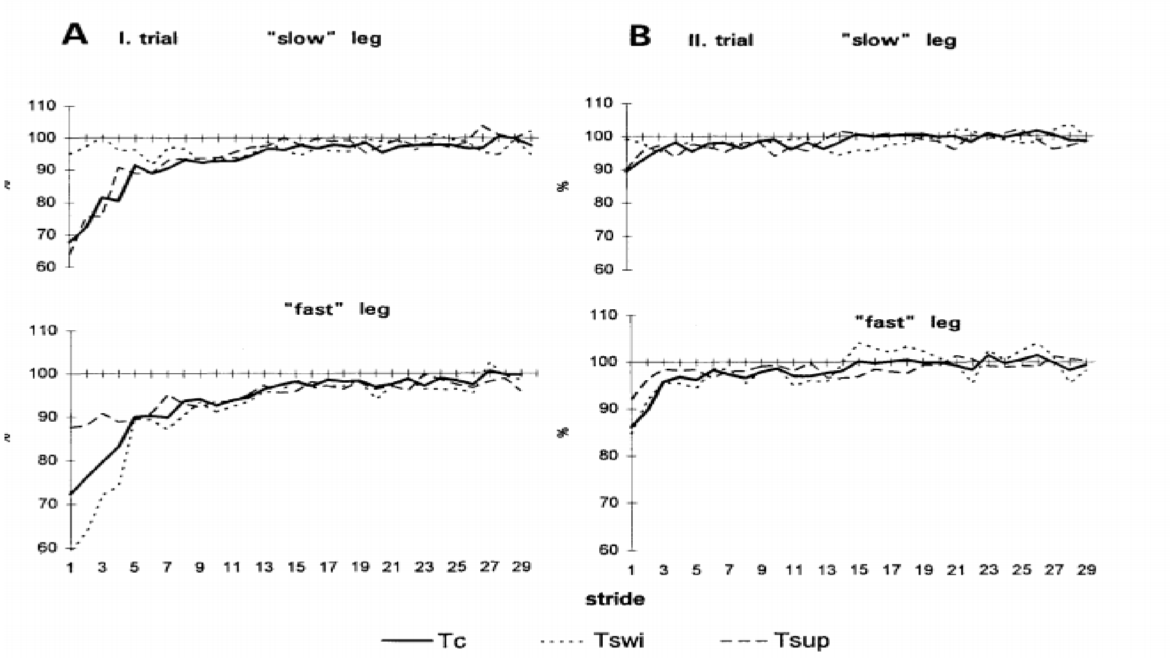


Figure 6. Stride cycle time (Tc), support time (Tsup), and swing time (Tswi) versus successive strides after the onset of split-belt walking as group means of all subjects (n= 11). A) First trial B) second trial. The values are expressed as a percentage of their respective adapted mean values, therefore 100% would equate to baseline. Note the more rapid adaption time to baseline in the 2nd trial exposure. *Figure adapted from Prokop et al. (1994)*

Reisman and colleagues (2009) continued this work and found that the adaption transfers to over ground walking in post-stroke patients. The extent to which this process is clinically relevant was further emphasised by a long term program of 30 minutes of split belt training for 3 day/week for 4 weeks that resulted in long term (up to 3 months) improvements in gait for some patients (Reisman et al., 2013). The initial asymmetry must have been over 5cm and there were considered to be responders and non-responders.

The effects of SB walking compared to overground walking have been investigated. It was found that even with a small gap (4 mm) between the belts results in a comparably large change in step width of 3.7cm to accommodate. This in turn resulted in increased hip adduction and valgus at the knee (Altman et al., 2011). It could be assumed that this response is to prevent trips as a result of crossover of feet on belts travelling at different speeds. Although not tested it could be hypothesised that the higher the speed difference, the greater the deviation from normal walking gait the subjects will display.

In order to fully appreicate how gait is adapted to the changes in envornment, there must be a method of quantification. This is provided in the form of motion capture.

Motion Capture

The study of human movement is evident throughout the ages in art and sculpture. With the development of photography attempts were made to capture the nature of motion (Muybridge, 1880s). In the 1980s Prof. John Paul made great leaps in motion capture through use of multiple video cameras recording human movement. Small markers, highly reflective to infrared light were attached to specific points on the body. Using complex mathematics, the position of these markers in space could be mapped throughout the movement. This principle provides the grounding for 3D motion capture as it is known at present (Cappozzo, 2015).

Current Vicon motion capture (Vicon Motion Systems Ltd., Oxford, UK) is an advanced version of this method. Small reflective markers are placed on anatomical landmarks on the body. These provide a guide for the skeletal positioning of the body as it moves. In reality the human body is immensely complex and a full reconstruction of its movements is not possible. Attempts have therefore been made to model this in attempt to simplify it. There are a number of models however the most widely used is the Plug in Gait (PiG). It offers a simplified segmental modelling system based on assigning complex anatomy to geometric shapes and joints. Creation of triangular segments allows position and orientation to be captured. It divides the lower body into seven segments; the pelvis and bilateral femur, tibia and foot. Experimental set up is quick and easy with minimal invasiveness and offers the ability to get real time feedback on body positioning.

Marker misplacement is the most common source of error in gait analysis, this is due to anatomical landmarks being difficult to palpate and replicate (Gorton et al. 2009). Small differences in marker placement can cause large error; this is particularly evident in measurement of abd/adduction and internal/external rotation.

There are a number of proposed alternatives to the PiG model, Duffell et al. (2014) suggests a cluster system that does not rely as heavily on identification of bony landmarks for joint centre positioning, in theory this is an improvement to PiG, however, it shows comparable performance when tested (Collins et al., 2009). In order to test accuracy and validity there must be a standard measure to which it can be compared. Unfortunately at present there is no recognised gold standard measurement system. Systems such as open CT and MRI machines may prove otherwise, however, they are vastly more expensive and spatially limiting. In the absence of this gold standard the main validation of gait models comes with comparison against previously collected data. The PiG model has been so widely used there is extensive data to which it can be compared. This is the model's most valuable asset, and while the system is by no means faultless it can be regarded as the most widely validated model. Baker (2013) concluded that until a strong evidence based validation of these alternatives the PiG model should be used.

Motion analysis may be carried out in an open lab in which subjects walk down a walkway into a specific area of capture. As emphasised previously treadmills provide an environment where a large number of stride cycles can be repeated in a small area. Integration of a treadmill into a motion capture space would therefore be useful for both research and rehabilitation purposes.

Computer Assisted Rehabilitation Environment (CAREN)

The Computer Assisted Rehabilitation Environment (CAREN) extended system (MotekForceLink, Amsterdam, Netherlands) has been widely used as a research tool investigating rehabilitation and neuromuscular control (Kerr et al., 2014). The CAREN system is a high end, custom built suite that integrates biomechanics research with rehabilitation training. It features a circular 3m diameter platform with a dual belt instrumented treadmill and mounted rig that supports a harness, ensuring safety for the subject. It is supported by hydraulic motion base capable of 6 degrees of freedom (DoF). This allows for huge versatility in its use, such as walking on a gradient, simulating trips through perturbations and importantly for this proposed research, split belt speeds.

It features 12 fixed mounted Vicon Bonita B-10 cameras that capture real-time motion capture. The platform is surrounded by a 180^{o} cylindrical screen projection system, featuring 3 projectors. A 4^{th} projector is positioned to display the chosen environment on the front portion of the platform floor to provide improved visual integration. The system utilises D-flow, a control software suite that provides real-time data streams between many types of integrated hardware. The software allows for control of the elements, the programming of which can be altered to suit the experimental protocol desired, although a range of standard applications are available. The system also features a full surround sound system to create a full multi-sensory environment.

Summary

Stroke is a major issue in the modern world and its impact on society through long term disability is vast. While advancements in rehabilitation methods have improved patient outcome a large number of patients are still faced with difficulty in walking a number of months and years post stroke. Gait is a seemingly simple, intuitive movement, yet it is associated with complex motor control pathways of a number of regions in the brain. Motor learning plays a vital role in the accommodation of new skills, repetition is linked to learning and so the ability to carry out a large number of steps in a controlled space is useful. A treadmill can allow for this and they have been shown to improve the rehabilitation of stroke patients.

Asymmetry is often a feature of a stroke patient’s gait, this is due to hemiparesis of one side of the body. Conventional treadmills, however, do not have the capability to tackle this aspect of rehabilitation. Recent studies have investigated the use of split belt treadmills in rehabilitation. The use of motion capture is an essential tool in the study of movement, novel technology has allowed for the integration of split belt treadmills and motion capture. This present study seeks to investigate the effect of split belt treadmills on gait and what role this may have in motor learning for stroke patients in improving asymmetry.

Methodology

The study was a controlled motion capture experiment in which spatial temporal parameters of gait in healthy individuals were recorded pre and post intervention. The intervention was in the form of exposure to SB treadmill walking. The Motek CAREN system was used for the experiment; this consisted of a circular platform with integrated split belt treadmill and 180^{o} immersive projection screen. It featured 12 Vicon B-10 motion capture cameras on a custom built permanent rig.

Participant Selection

This study was approved by the University of Strathclyde Ethics Committee. The study took place in the Biomechanics Lab in the Biomedical Engineering department of the Strathclyde University. Participants in the study were young, healthy volunteers recruited from peers across the University of Strathclyde. Prior to participant inclusion, the experimental process was explained in detail with any questions answered, emphasising that inclusion was voluntary and would be allowed to leave at any time. If the participants were comfortable to continue informed consent was taken. A sample participant information sheet & consent form can be seen in Appendix I.

Data from 10 subjects in total was recorded (7 males, 3 females), demographic data shown in Fig 7. All participants were recruited by email or personal correspondence using the following inclusion criteria; over 18 years old, in good health and naïve to SBT walking prior to experimentation. Subjects were excluded under the following criteria; individuals with known issues with walking 3 weeks prior to experimentation, individuals with a known condition during or prior to the investigation period or were taking any medication that may compromise their ability to participate. Further excluding conditions included pregnancy, current ongoing illness, illness within two weeks of session date, exercise intolerance, musculoskeletal surgery (e.g. hip replacement), neurological deficit, vestibular (balance, including travel sickness) problems, and sensory limitations not corrected with glasses or hearing aids.

Age	BMI	Walking speed (m/s)
23.2 ± 0.7	24.8 ± 2.5	1.14 ± 0.17

Figure 7. Demographic data of subjects recruited (n=10). BMI: Body Mass Index (Mean ± SD)

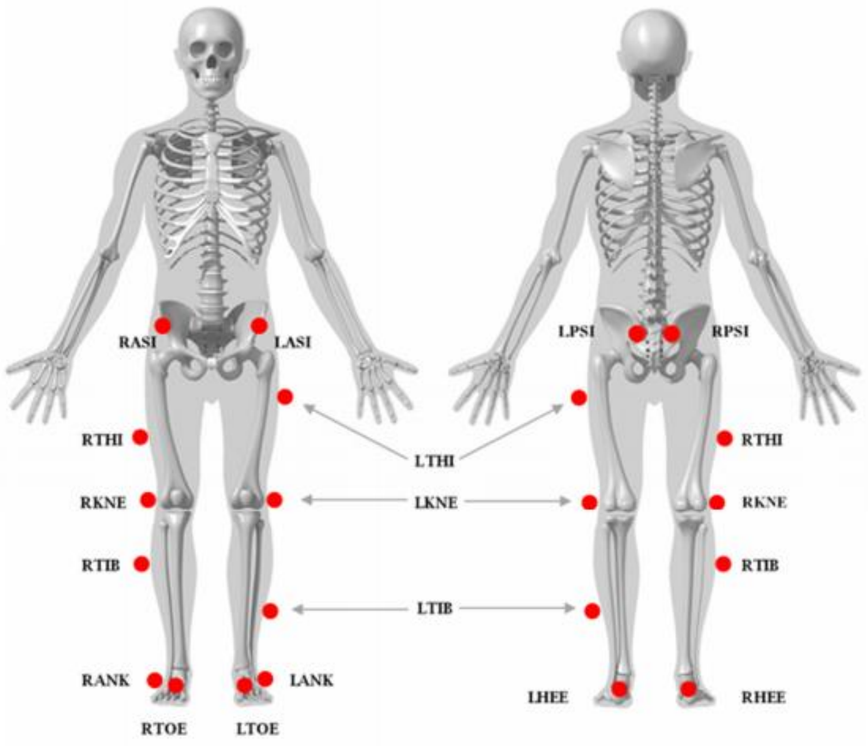
System Calibration

Calibration of the CAREN system is carried out following manufacturer protocol (Vicon, 2015). A precise active calibration wand was used to calibrate the Vicon cameras and set the ground of the system. The active wand features infrared LEDs with strobe setting at specific points of known length. This was performed at the start of each test day, as the environment is highly controlled it was not necessary to re-celibate between participants. Calibration provided a residual score as a measure of accuracy.

Data Capture

The study sought to record kinematic data involved in gait. To achieve this, lower body Plug in Gait (PiG) model was chosen, the supporting argument for this is laid out in the literature review. In short, the PiG model is simple and effective for recording kinematic data with extensive literature for comparison. Specialist reflective markers were laid out as instructed in Vicon PiG manual (Vicon, 2015) shown in Fig. 8. Subjects wore tight fitting lycra clothing, allowing for accurate marker placement, ensuring that marker position would reflect the movement of the underlying segment.

A



B

Marker Anatomical Landmark	Positioning (Symmetrical unless specified)
Left Anterior Superior Iliac Spine (ASIS)	Placed directly over the left anterior superior iliac spine
Left Posterior Superior Iliac Spine (PSIS)	Placed directly over the left posterior superior iliac spine
Left knee (KNE)	Placed on the lateral epicondyle of the left knee
Left thigh (THI)	Place the marker over the upper lateral 1/3 of the thigh, below the swing of the hand, although height is not critical (Right is placed on lower 2/3 of thigh)
Left ankle (ANK)	Placed on the lateral malleolus
Left tibia (TIB)	Similar to the thigh markers, these are placed over the lower 1/3 of the shank (Right is placed on upper 1/3 of shank)
Left toe (TOE)	Placed over the distal 1st phalanx
Left heel (HEE)	Placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker

Figure 8. A) Diagrammatic representation of the markers placement on anatomical landmarks. B) Describes in detail the position of the markers placement. (Prefix R=Right L=Left)

Participant Preparation

Subject demographic data were recorded as follows; height and weight using a stadiometer with integrated scales. To acquire the perceived dominant leg the subject was asked which foot they would most commonly kick a ball with. (Schneiders et al., 2010).

Right and left true leg length was recorded from ASIS to medial malleolus of the corresponding ankle using measuring tape. A pair of callipers, accurate to 1mm, was used to calculate inter ASIS distance, knee and ankle width. These recordings were input to the PiG model.

Experimental Protocol

Experimental protocol (Fig. 9) dictated that each subject would undergo a total of 15 minutes SBT walking in the form of two blocks; 10 and 5 minutes, separated by a washout in the form of time walking with belts tied at a slow speed and a balance game.

Qualitative data in the form of observation of general trends, participant perception of walking and number of trips were recorded on a data sheet (Appendix II).

Normal Walking Speed Calculation

Normal comfortable walking speed was ascertained by gradually increasing belt speed until the subject declared a comfortable pace was reached. The speed was then increased to what is considered an uncomfortable pace. Belt speed is then reduced until the subject again declares a comfortable walking speed. The average of the two declared comfortable walking speeds is then taken as the normal walking speed.

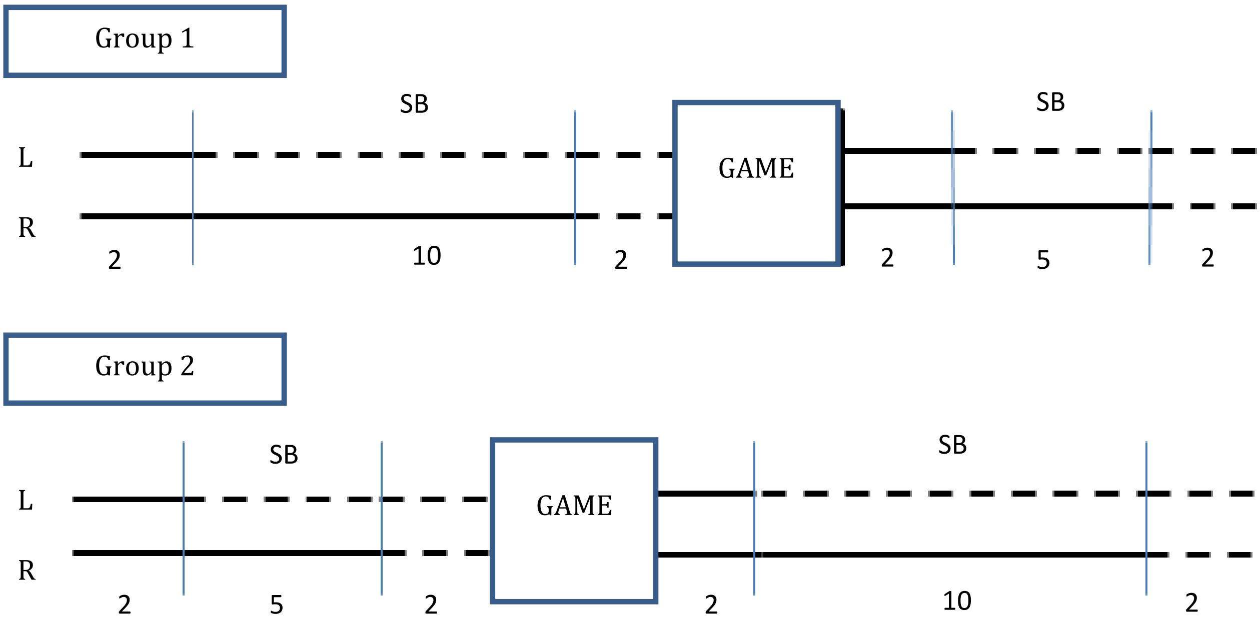


Figure 9. Example of experimental protocol (R=right leg; L=left leg), solid line indicates fast treadmill speed and dashed line slow speed (2:1 ratio). Participants were placed in either group 1 or 2. Both underwent a total of 15 minutes SB walking. Subjects underwent 2 minutes of normal walking followed by block of SB walking (5/10 minutes) and 2 minute return to tied slow walking speed (1:1). (approx. 2 minutes). Dominant leg is placed on fast belt (as right dominance is most common this is given in example).

Split Belt

Subjects underwent a period of acclimatisation to the CAREN environment. The screen displayed an immersive computer generated forest pathway scene that was on an endless loop (Fig. 10). Subject walked at the self-selected normal speed with belts tied for 2 minutes. After this period the subject began the first block of SBT walking. The previously ascertained dominant foot was assigned to be the "fast leg" and consequently the non-dominant leg was the "slow leg". The fast leg was maintained at the normal walking speed while the slow leg belt speed would be reduced to half the value of the normal walking speed. This continued for the allotted time (10 or 5 minutes). At the end of the SBT walking the belts were returned to the tied condition of the slow speed. There then followed a 2 minute period of re-adaption to normal treadmill walking speeds. Treadmill acceleration/deceleration was set to the manufacturer default, based on years of experience and feedback, of 0.25m/s, providing a comfortable but rapid transition to different belt speeds.

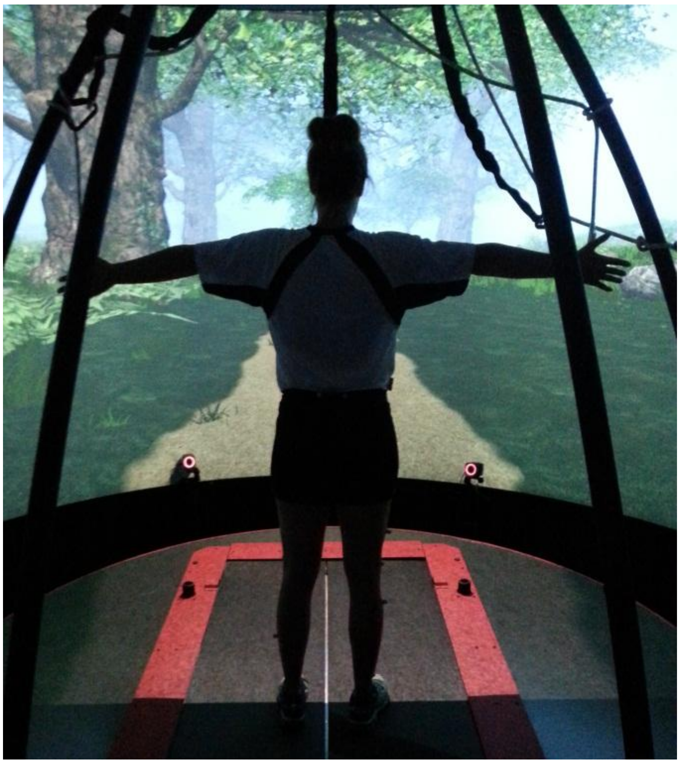


Figure 10. Photograph of subject during calibration, prior to initiation of walking. The forest walk projection can be observed along with split belt and safety harness set up.

Game

The balance game was in the form of a car control exercise in which the aim was to get as far as possible down a street while avoiding cars coming towards them in the opposite direction. Shifting body weight accelerated the car and moved it left and right. The game lasted 120 seconds and had no purpose other than to offer a break to the participants and attempt washout the effects of SB walking.

Data Processing

Data were stored on University computers. The files were viewed in manufacturer software (Nexus version 2.1.1) and cropped to a desired size. This was approximately 20 stride cycles in transition periods and 5 stride cycles in baseline and intermittent readings. Key events were labelled in the sequence by position of markers to define toe off and heel strike for each foot. Heel strike was defined as the lowest dip in sagittal plane trajectory of the heel marker. Toe off was defined as the point at which the toe marker had a velocity of 0 m/s in the transverse plane, immediately prior to acceleration of toe off.

Cropped trials were processed through a pipeline, producing an output excel file containing data. Kinematic data of all lower body segments were outputted directly by the PiG model. Step length was defined as the distance between the point of initial contact of one foot (heel marker) and the corresponding distance to the opposite toe marker.

Step length symmetry was calculated as follows:

$$\text{Step length symmetry} = \frac{Fast\ leg\ step\ length\ (m) - Slow\ legstep\ length\ (m)}{Fast\ leg\ step\ length\ (m) + \ Slow\ legstep\ length\ (m)}$$

Microsoft Excel 2010 was used to compile data and create graphs.

Statistical Analysis

Hypothesis was tested using graphical interpretation. Paired t-tests were used to test differences of asymmetry in means of first consecutive strides after transition.

Results

All participants completed the experiment following the protocol and as such data from 10 participants is presented below. All subjects were able to cope with the novel introduction to split belt (SB) walking.

On initiation of SB walking there was reported some difficulty but this was tolerable and there were no falls. There were, however, a number of trips and slips recorded for each individual that could be accounted to partial crossover of feet on the belts when traveling at different speeds. This was particularly evident during the transition periods where subjects appeared off balance. All subjects reported increased attention required coping with SB condition, however, after a period all subjects reported a perceived adaption in which SB walking felt comfortable. This was shown further by a perceived altered sensation when returning to normal tied walking conditions, where one leg was considered to be traveling faster than the other while traveling at the same speed. Upon second exposure to SB all subjects self-reported perceived adaption in a shortened time.

Step Length Asymmetry

An unexpected observation was found in that subjects displayed a different response in relation to SB walking (Figure 11). The majority of subjects (n=6) showed a consistent pattern of results and these will be discussed in detail. This response was negative asymmetry in response to SB conditions during early adaption (EA) (Response 1). This implies that the fast leg step length is shortened relative to the slow leg. Conversely during post adaption (PA) the fast leg showed a relatively shortened step length. A proportion (n=4) of participants, however, displayed the opposite response, showing positive asymmetry during EA and negative asymmetry in PA (Response 2). The validity of these results is questionable, and may be due to experimental error, but for completeness of results these will be presented and considered.

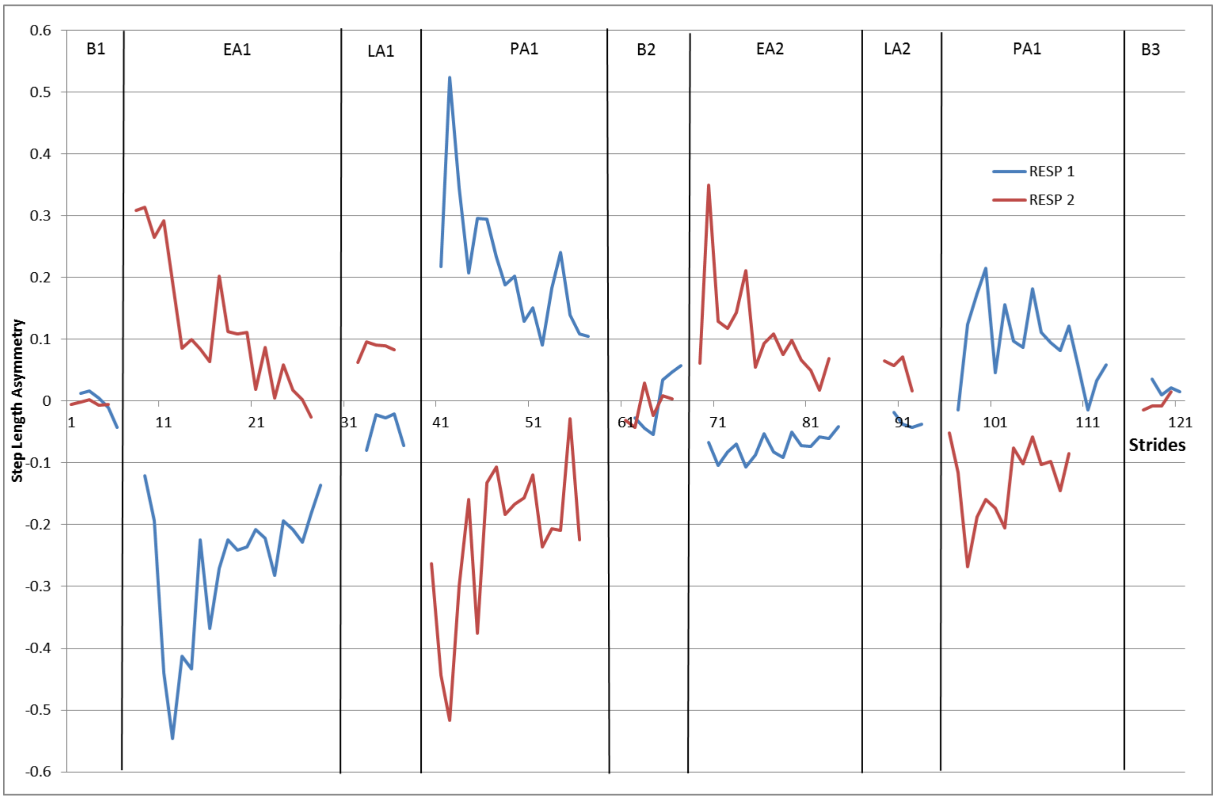


Figure 11. Displays the mirrored difference in response in asymmetry to SB walking exhibited by subjects. The first group (n=6) of responders (blue) show initial negative asymmetry in early adaption (EA) before positive asymmetry in post adaption (PA). The second group (n=4) show the mirrored response, i.e. the step length alterations occur on the opposite leg. B1: initial baseline, EA1: early adaption to first exposure of SB (first 15 strides after transition), LA1; Late adaption to first exposure of SB, PA1: Post adaption of first return to normal treadmill conditions (first 15 strides after transition); B2: baseline between blocks of SB session, EA2: early adaption to second exposure of SB (first 15 strides after transition), LA2; Late adaption to first exposure of SB, PA2: Post adaption of second return to normal treadmill conditions (first 15 strides after transition), B3: final baseline reading.

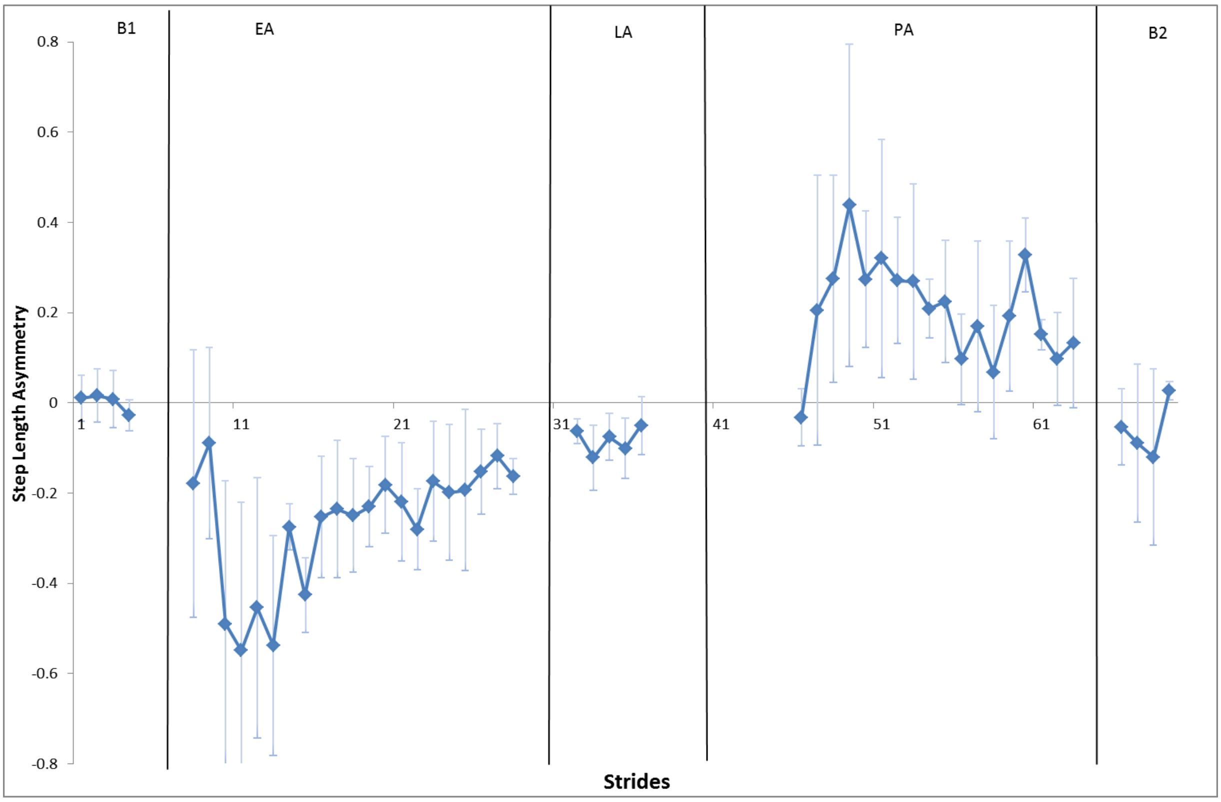


Figure 12. Mean step length asymmetry trends during consecutive strides in experimental protocol (n=6). Each data point represent a stride asymmetry. B1: initial baseline, EA: early adaption to first exposure of SB (first 15 strides after transition), LA; Late adaption to first exposure of SB, PA: Post adaption of first return to normal treadmill conditions (first 15 strides after transition); B2: baseline between blocks of SB session. Error bars show SD.

Figure 12 shows the mean values of step length asymmetry of participants over experimental protocol. The overall stride number does not correlate to total strides as there are large gaps in motion capture. All subjects showed baseline (B1) symmetry in step length (Mean: -0.002, SD: ±0.03). Exposure to SB treadmill walking resulted in a swing towards asymmetry as step lengths are altered to accommodate the condition. As consecutive stride cycles are recorded this asymmetry decreases towards the baseline levels, this can be considered the early adaption phase (EA). After prolonged exposure, late adaption (LA), the subject has altered their gait parameters to re-instate symmetry to accommodate SB walking. Upon return to normal treadmill walking conditions the initial asymmetry displayed in EA occurs in the opposite limb, this is termed post adaption (PA). Again this shows a trend of readapting towards the baseline over consecutive stride cycles. After a short period of normal walking the subjects return to baseline symmetry as shown B2.

Group 1 and Group 2 are shown together in Fig. 13. Observationally; the results for asymmetry show a similar trend. This shows that a difference of exposure time (10 minute block first (Group 1) or 5 minute block first (Group 2)) does not appear to impact on the PA asymmetry or the speed of adaption upon second exposure to SB.

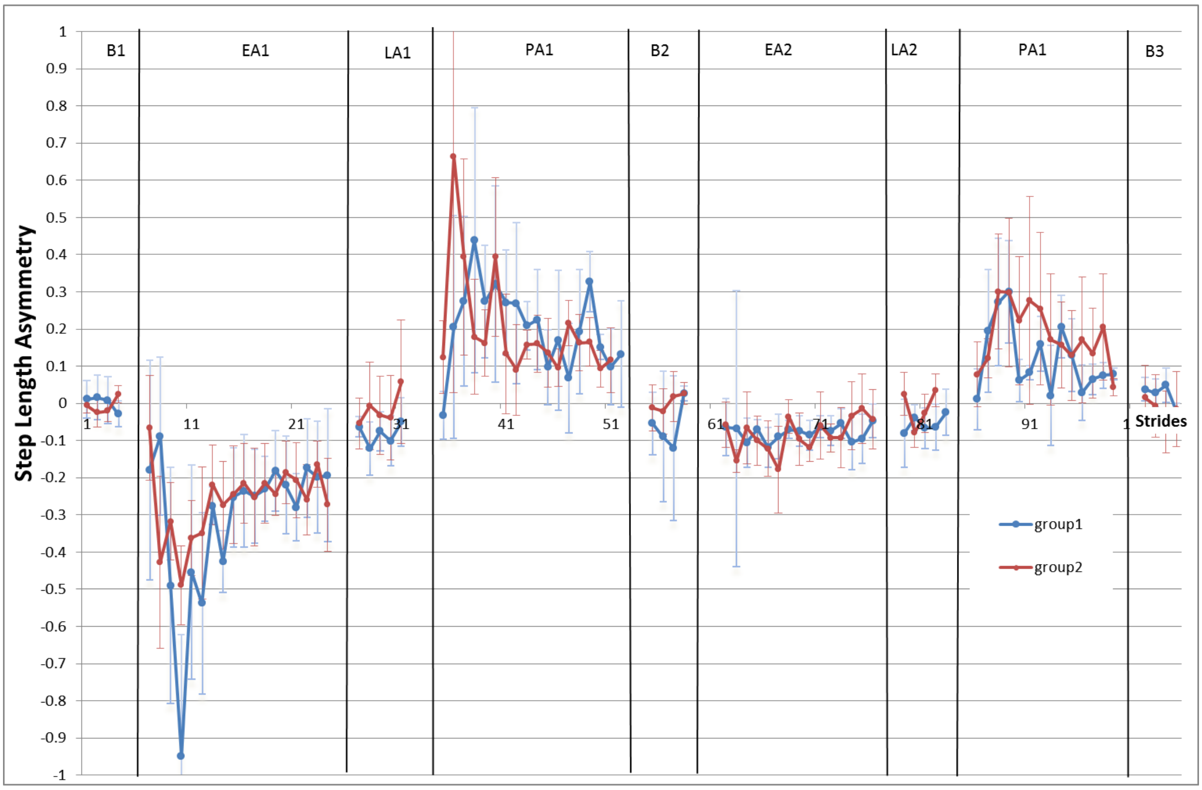


Figure 13. Step length asymmetry during consecutive strides in each motion capture block. Group 1 (blue) underwent a 10 minute block of SB first followed by a 5 minute block. Group 2 (red) underwent a 5 minute block of SB first followed by a 10 minute block. The groups show comparable trends. B1: initial baseline, EA1: early adaption to first exposure of SB (first 15 strides after transition), LA1; Late adaption to first exposure of SB, PA1: Post adaption of first return to normal treadmill conditions (first 15 strides after transition); B2: baseline between blocks of SB session, EA2: early adaption to second exposure of SB (first 15 strides after transition), LA2; Late adaption to first exposure of SB, PA2: Post adaption of second return to normal treadmill conditions (first 15 strides after transition), B3: final baseline reading**.**

Repeated exposure of SB walking again caused a swing towards asymmetry in the EA2 phase. The level of asymmetry, however, can be observed to be greatly reduced when compared to EA1 (Fig. 14). The results show accommodation adaption towards a baseline is faster upon repeated exposure to a greater extent than time of exposure, at least when considering relatively short exposure times of 5 and 10 minutes. LA2 similarly shows a trend towards baseline asymmetry. In PA2 there is again a trend towards opposite asymmetry, to a comparable level of that seen in PA1 (Fig. 15).

Step Length Asymmetry					
	1st Exposure	**SD**	**2nd Exposure**	**SD**	
B1	0.001	±0.01	-0.002	±0.03	
EA	-0.299	±0.09	**-0.085**	**±0.05**	*p<0.001
LA	-0.002	±0.01	-0.007	±0.01	
PA	0.206	±0.10	0.154	±0.06	
B2	-0.047	±0.01	**0.009**	**±0.01**	*p<0.05

Figure 14. Mean step length asymmetry in each block. 2nd exposure EA shows significantly less step length asymmetry over first 15 steps than 1st exposure (Paired T-test, $p<0.001$). A significant difference was also found in final baseline asymmetry. Due to small sample size significance must be treated with caution. B1: baseline, EA: early adaption, LA1; Late adaption, PA1: Post adaption; B2: end baseline

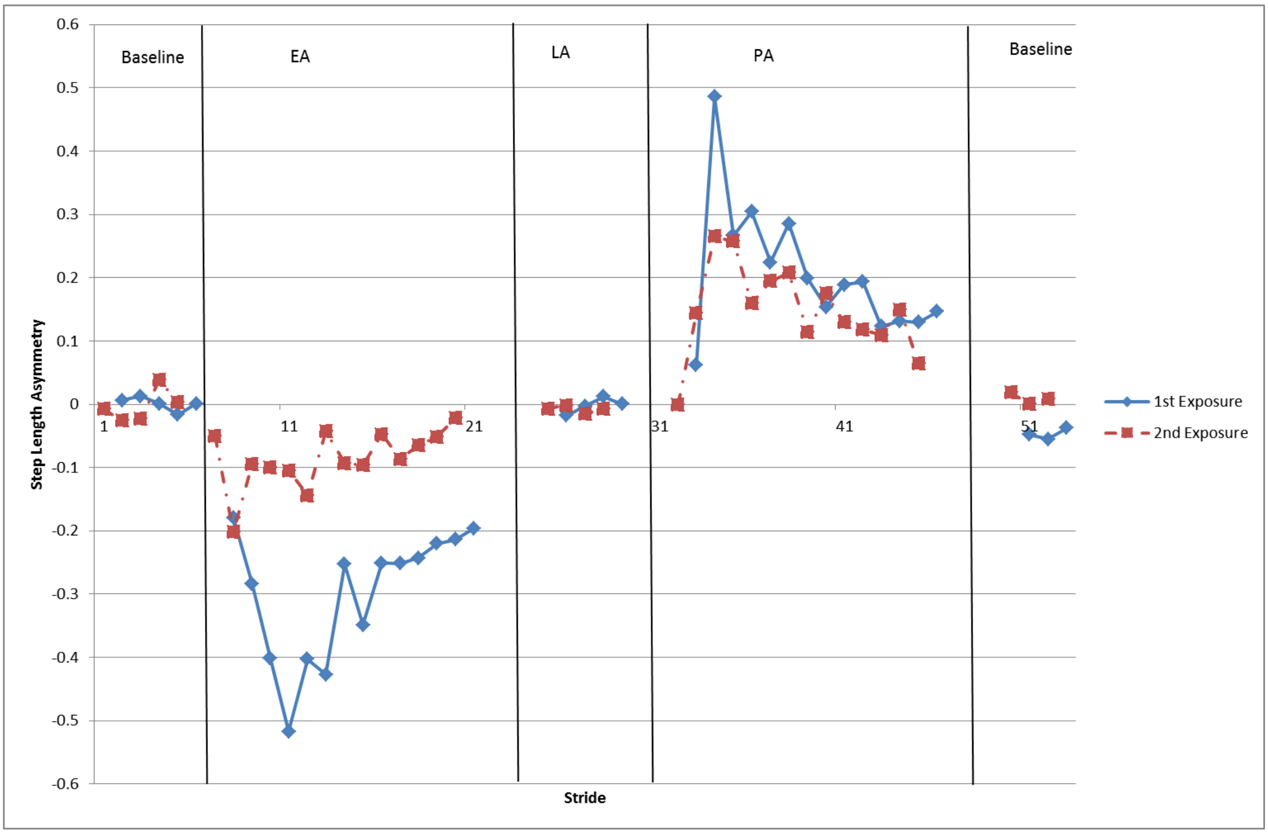


Figure 15. Mean Step length asymmetry during first exposure to SB (blue) and repeated exposure to SB (red), i.e. EA1 = blue line in EA. Asymmetry is greatly decreased during early adaption phase (EA) upon second exposure to SB compared to the first exposure. Baseline, late adaption (LA) and post adaption (PA) show similar trends.

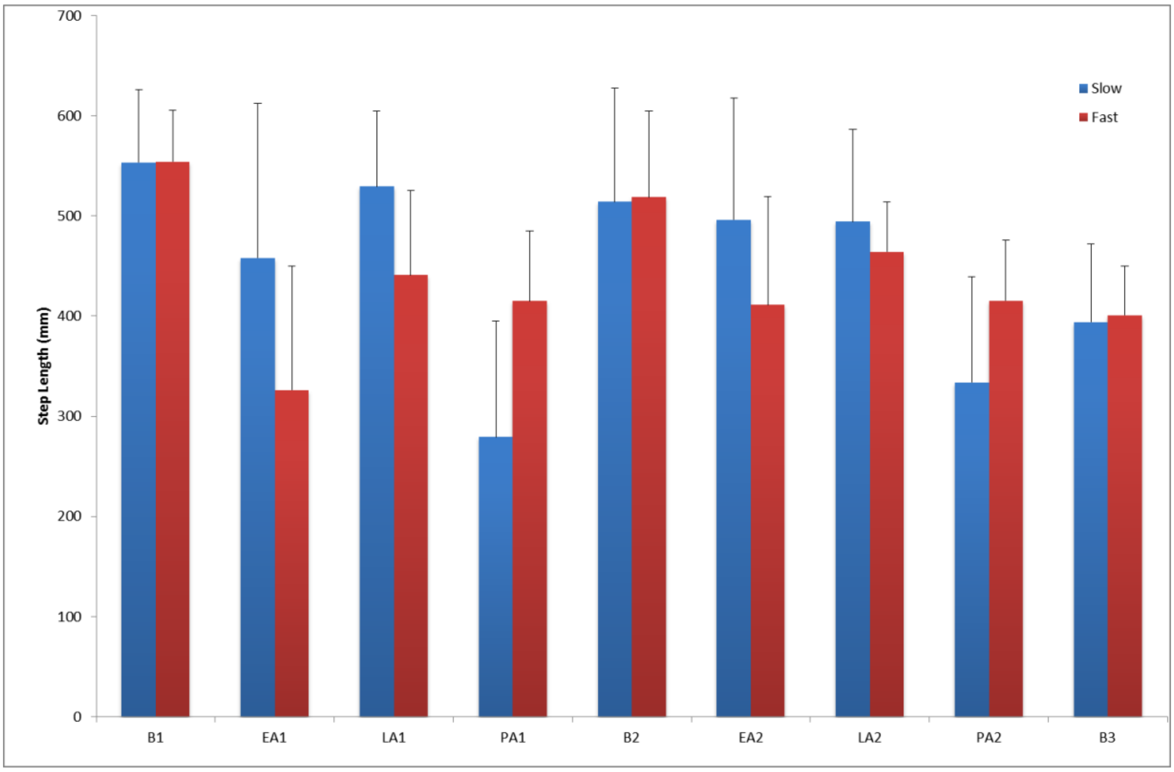


Figure 16 Absolute mean step lengths from each data capture of limb on fast (red) and slow (blue) belt. Fast step length decreases relative to slow step length during EA and the opposite can be observed in post adaption as fast step length is increased relative to the slow leg. It is also noted that overall step length appears to decrease during the transitional phases. B1: initial baseline, EA1: early adaption to first exposure of SB (first 15 strides after transition), LA1; Late adaption to first exposure of SB, PA1: Post adaption of first return to normal treadmill conditions (first 15 strides after transition); B2: baseline between blocks of SB session, EA2: early adaption to second exposure of SB (first 15 strides after transition), LA2; Late adaption to first exposure of SB, PA2: Post adaption of second return to normal treadmill conditions (first 15 strides after transition), B3: final baseline reading. (n=6)

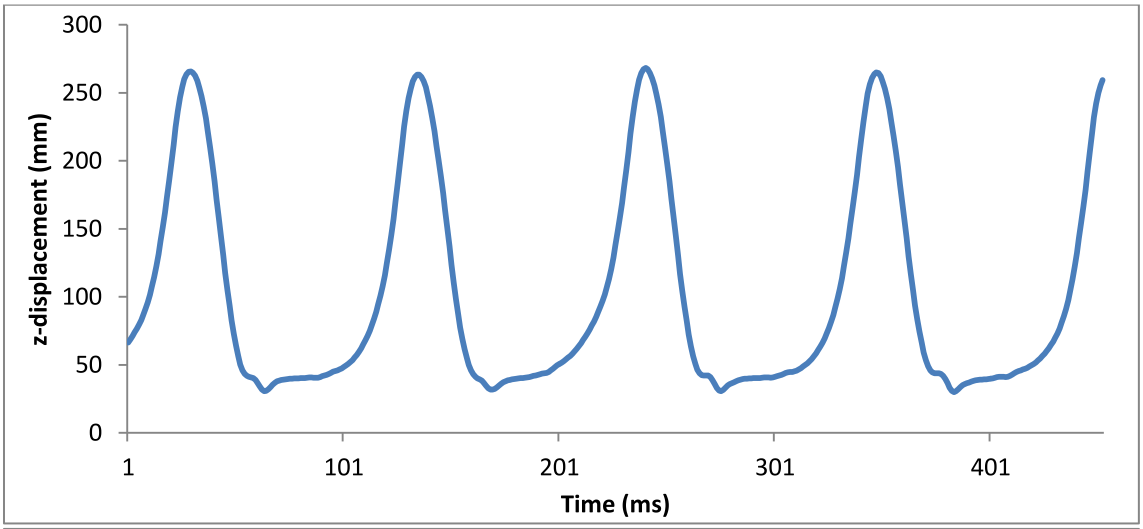
To summarise, the adaption to SB can be characterised by asymmetry in step length, as a result of alterations of step lengths of legs on fast and slow belt (Fig. 16). Initially, asymmetry is large, this is decreased as an adaption process tends towards a baseline. The asymmetry is reversed when returned to normal walking. The number of exposures, as opposed to length of time of exposure, appears to be the important factor in relation to speed of SB adaption.

Strategies

The strategy each individual adopted to accommodate the split belt gait was varied. One particularly common strategy was noted and will be explored. The foot on the fast belt, prior to heel strike, was held briefly in the air to extend the swing phase period. This will be referred to as the "heel hang" strategy. The adoption of this strategy can be seen in a case study by analysis of the heel marker position in the sagittal plane (z-axis) of a single participant. The baseline reading shows the normal trace of heel strike for an individual that features a distinct trough at heel strike as load is transferred to the heel, this is consistent for all baseline conditions (Fig. 17 A). During the first EA phase (Fig. 17 B) there is high variation in the spacing of the heel strikes and shows no distinct pattern, this is reflected in the high asymmetry experienced. By the 13th stride, a second peak can be observed prior to heel strike (denoted by *). This shows that the heel is being held at full knee extension and hip flexion to prolong the swing phase of the fast limb. The pattern is not consistent and fades before returning, highlighting the variability.

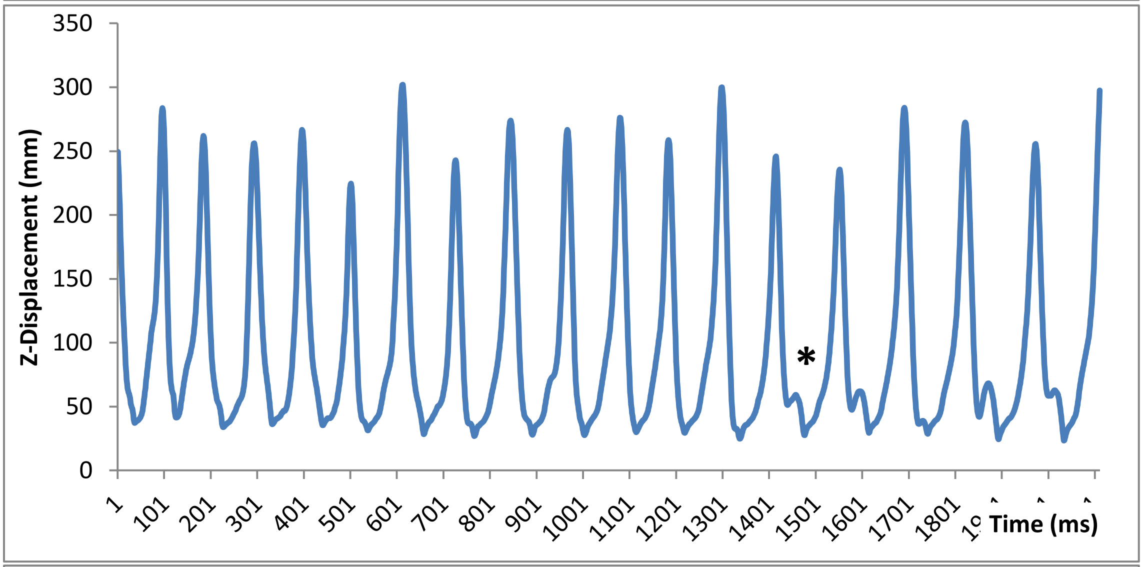
During LA the distinct and repetitive pattern can be seen as a second peak in each step (Fig. 17 C). This shows that the heel hanging strategy has been adopted by the subject and correlates with the return towards baseline symmetry. In the second exposure to SB walking the adoption of the heel hang strategy is more rapid, by the 3rd step the trend can be observed (*) that is relatively consistent throughout the rest of the EA phase (Fig. 17 D). This corroborates the asymmetry data showing less of a deviation after second exposure and also the participant's verbal reporting that the transition is more easily accommodated. The final baseline recording shows may be perceived as an artefact of this strategy persisting into normal baseline gait (Fig. 17 E).

A)



z-displacement (mm)
300
250
200
150
100
50
0
1
101
201
301
401
Time (ms)

B)



Z-Displacement (mm)
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101
201
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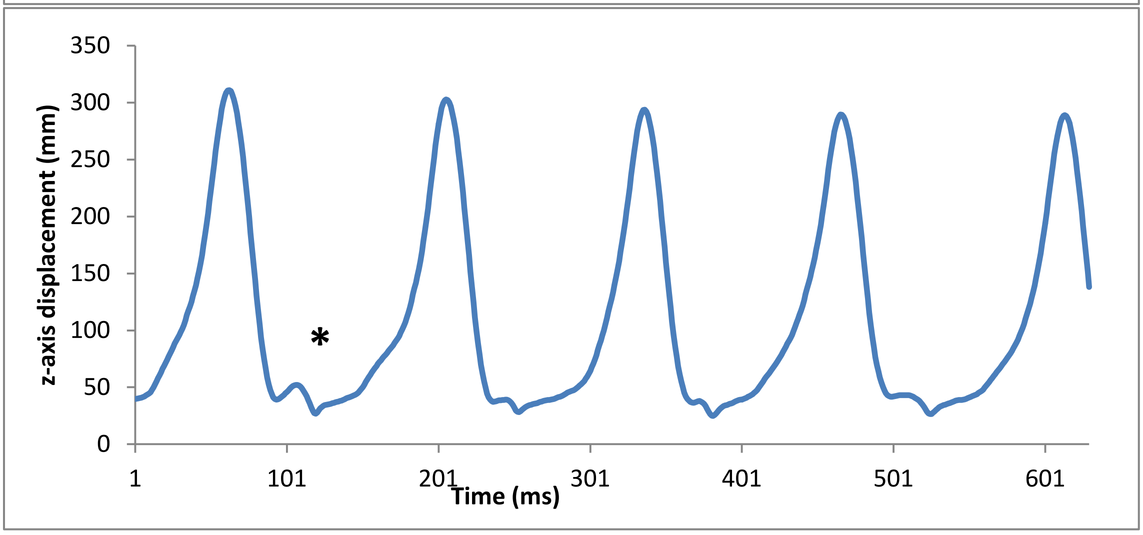
1801
Time (ms)

C)



z-axis displacement (mm)
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1
101
201
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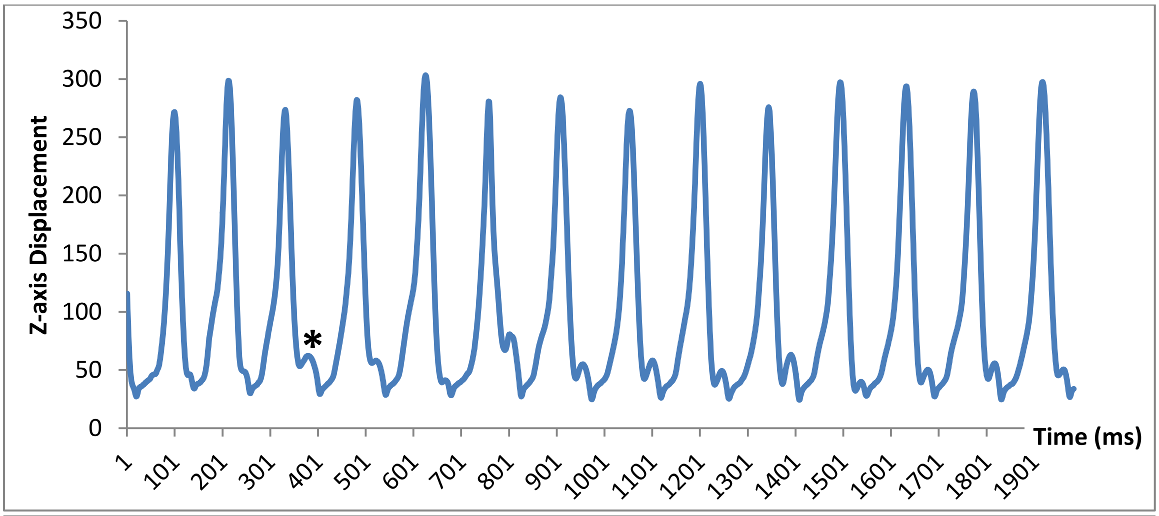
401
501
601
Time (ms)

D)



E)

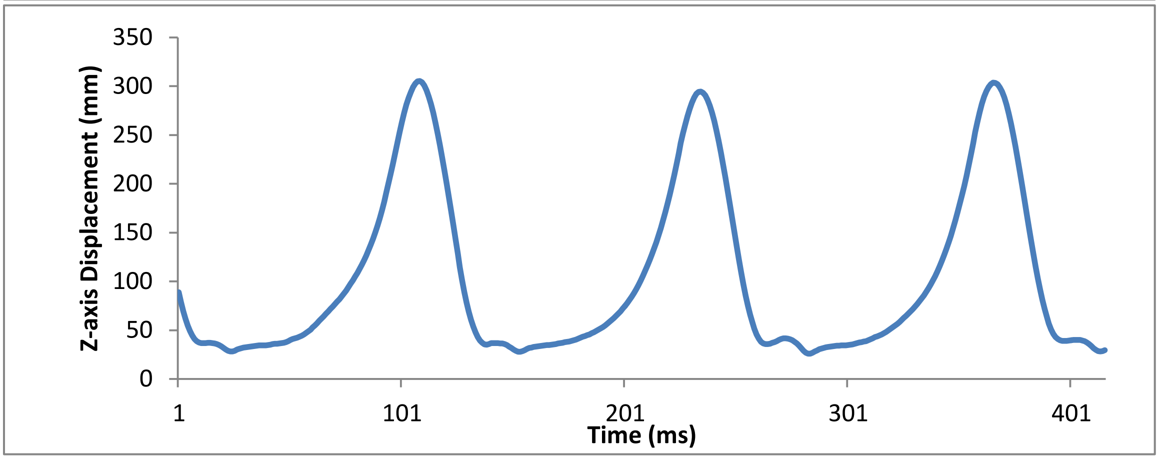


Figure 17. Displacement of the right foot heel marker in the z-axis (transverse plane). A) Baseline trend. B) Early adaption (EA) on first exposure to SB. * denotes adaption behaviour termed "heel hang" occurring during the 13th step after transition to SB walking. C) Late adaption (LA), shows the heel has been adopted consistently. D) Early adaption (EA) on second exposure to SB. * denotes obviously heel hang adaption occurring during the 3rd step after transition to SB walking. E) End baseline reading after SB exposure, a shadow of the heel hang strategy can be observed.

To summarise, a distinct strategy of accommodation to SB was identified. This was in the form of a prolonged swing phase of the fast limb by use of heel hang strategy, thus increasing step length. This evidence provides a strategy for the trend of asymmetry shown previously. Similarly the strategy can also be observed to have a more rapid uptake upon repeated exposure and result in an artefact that persists in post SB walking baseline gait. The precise number of strides taken to accommodate could not be confidently stated due to the variable nature each individuals trend towards a symmetrical gait.

Figure 18 shows step length data for the group that responded directly opposite to the majority of subjects. The reasons for this are not obvious and may be the result of error in data handling. Similar to the first group however these subjects exhibited a swing towards asymmetry that was reversed when returned to normal walking. It should be noted also that both groups display a reduced overall step length during transition periods.

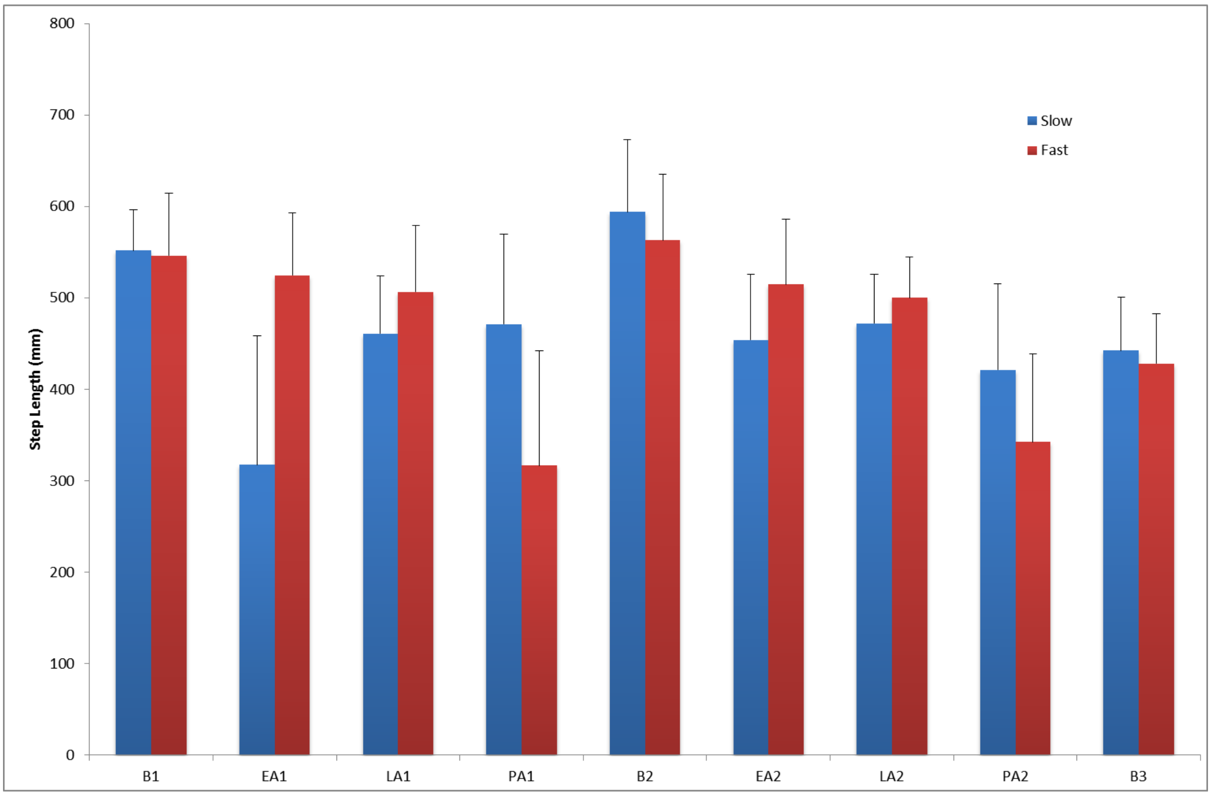


Figure 18 Mean step lengths from each data capture of limb on fast (red) and slow (blue) belt. This group responded in an unexpected, opposite way to the majority of subjects; showing a decrease in slow leg step length in early adaption and the opposite during post adaption. B1: initial baseline, EA1: early adaption to first exposure of SB (first 15 strides after transition), LA1; Late adaption to first exposure of SB, PA1: Post adaption of first return to normal treadmill conditions (first 15 strides after transition); B2: baseline between blocks of SB session, EA2: early adaption to second exposure of SB (first 15 strides after transition), LA2; Late adaption to first exposure of SB, PA2: Post adaption of second return to normal treadmill conditions (first 15 strides after transition), B3: final baseline reading. (n=4)

Discussion

The aim of this study was to investigate the gait adaptions that occur in healthy subjects when exposed to SB walking. It sought to characterise this adaption through use of motion capture and investigate the effect of duration and repetition of exposure.

It was shown that upon exposure to split belt (SB) treadmill walking, healthy young adults were able to adapt their gait. SB walking created an initial asymmetry in step length between the right and left leg. This asymmetry subsequently decreased towards the baseline levels, constituting an early adaption phase (EA). A late adaption (LA), taken as measurement towards the end of each SB session, was characterised by the return to a baseline level as the individual accommodated their gait to regain symmetry, a key target in the normal gait pattern. After exposure to the prescribed period of SB walking, the subjects were returned to normal treadmill walking with belts tied, termed post adaption (PA). This resulted in a reversal of the asymmetry displayed in EA. Here, step length of the opposite limb shortened. Again this asymmetry tends towards baseline as the subject continues to walk under normal conditions. Upon subsequent exposure to SB walking subjects showed significantly less asymmetry ($p<0.001$) during EA, reaching a baseline asymmetry in a reduced number of strides. When returned to normal conditions, however, there was no significant difference in the PA asymmetry ($p = 0.975$). The time spent in SB walking did not result in differences in PA asymmetry as may have been expected in a dose response reaction.

An unexpected outcome was shown as two distinct responses to SB walking. One group (n=6) was shown to shorten the step on the slow belt during EA and the other shortened the step on the fast belt (n=4). Both groups showed reversed asymmetry in PA and overall displayed a mirrored response to each other. The reason for this is not obvious. It could be postulated that individuals may tend towards a preference in which step length is adapted, if this were found to be true this would be a significant finding that has not been reported previously. The recruitment of this response, however, is difficult to accept as it is counter-intuitive. Shortening of the step on the slow belt would likely cause further instability and would possibly result in a double stepping pattern, that was not observed. The most likely scenario then is that there has been an error in the computational analysis of the data although the source of this has not been identified. To fully test this error, a repeated measure of the specific subjects following repeated experimental protocol would yield a definitive result, either highlighting an unreported trend or confirming a computational error.

During the adaption period a distinct strategy was noted, this has been termed heel hang. During adaption to SB it was noticed that heel strike was being delayed slightly to prolong the swing phase of the fast leg, thus increasing relative step length as the standing leg on the contralateral side recedes by action of the belt. This was displayed by the heel marker positioning in the sagittal plane as a distinct peak prior to heel strike that differed from baseline recorded in individuals. In the case study of one participant this was shown to be adapted by the 13th stride after transition. Upon repeated exposure to SB this adaption occurred by only the 3rd step. This strategy was observed during testing on a number of subjects and further data analysis is required to determine if this is a typical response.

This study builds on work on SB walking in healthy adults (Prokop et al., 1995; Reisman et al., 2005; Huynh et al., 2014). In terms of qualitative feedback, similar to previous studies, subjects reported perceptual after-effects in which they perceived the limbs were traveling at different speeds even though they were not. This is considered by previous studies to be due to the change in expected vs. actual proprioceptive feedback from the movement. Previous studies reported that during initial adaption to SB walking, the limb walking on the slow belt exhibits a greater step length (Reisman et al., 2005; Huynh et al., 2014). In agreement with previous studies it was shown that subjects underwent an adaption process during prolonged exposure to SB towards a baseline level of symmetry, termed late adaption (LA). While previous studies reported the average asymmetry values of the first 5 strides, this study showed the progression of asymmetry in the first 15 consecutive steps of SB walking, displaying the trend towards adaption. Reisman et al., (2005) reported adaption to a 2:1 SB within approximately 20-30 strides. Results presented suggest a slightly faster adaption process, however, the precise definition of adaption is somewhat ambiguous due to the variability of step length under these conditions. When returned to normal walking the asymmetry is the reverse of that displayed in early adaption to SB, previous studies showed this resulted in a longer step length occurring on the limb previously on the fast belt (Prokop et al., 1995; Reisman et al., 2005; Huynh et al., 2014). The level of asymmetry experienced in PA is slightly lower than those reported by Huynh et al (2014) but as stated this was taken as the mean of the first 5 strides and it is liable to high variability.

It was noted in this study that overall step length generally decreased as a result of SB walking with subjects often taking smaller steps than at baseline. This was not reported in previous studies with only a single leg showing reduced step length at each transition (Huynh et al., 2014). This is most likely due to the slower walking speed at final baseline recording.

Both Reisman et al. (2005) halted the belt between transition to SB and normal treadmill conditions and Huynh et al., (2014) preformed over ground walking tests after SB conditions. This may somewhat alter the adaption process as it is not a continuous flow in walking gait and over ground walking may create significant differences in gait post adaption. Further to this subjects in the present experiment did not have any form of support in the form of front or side bars, differing from other reports (Reisman et al., 2005; Prokop et al., 1995; Dietz et al., 1994, Huynh et al,. 2014). This may be important as the use of supporting bars improves balance and stability and perception of risk of fall

(Jeka and Lackner, 1994). As symmetry is important in terms of overall stability when walking, the use of additional support may reduce the need to regain symmetry. In this as subjects were offered no support, other than a safety harness to prevent falls, as such there may have been more emphasis for an individual to regain symmetry to improve balance. This may account for the slightly shortened adaption process that was displayed by participants in the present study.

SB walking causes an initial asymmetry in step length. This is due to one limb travelling faster than the other, thus normal walking pattern is unable to accommodate for this. The neuromuscular control system is able to adapt the individual's gait to restore symmetry; this is considered to be an action of the cerebellum as a result of feedback and feed forward motor control (Morton & Bastian, 2006). When returned to normal walking conditions the asymmetry was pronounced on the opposing side to a similar extent. This has been termed the "aftereffect" or post adaption, it is considered to be the result of an overcorrection as a result of the adaption to altered gait that was learnt during exposure to SB (Reisman et al., 2005; Reisman et al., 2007).

Upon repeated exposure, the asymmetry experienced during early exposure was significantly reduced. This could be attributed to a degree of motor learning. It was shown that a distinct strategy can be adopted to accommodate SB walking. This strategy was picked up in a reduced number of strides after second exposure. The more rapid accommodation can be attributed to a learned event. When returned to normal walking, however, after the second exposure to SB walking there was no significant difference in opposing asymmetry (post adaption). This could be considered to be related to the fact that normal walking is a motor skill that is a deep-rooted action in healthy subjects. Therefore, the adaption process back towards a normal walking gait, for healthy subjects at least, could be considered to be constant. This may not be true for those with difficulties in gait or undergoing a prolonged period of gait re-training exercises as a process of motor learning could take place. Further work by Reisman et al., (2013) has shown that after extensive, repeated training of post stroke patients on SB treadmills results in improvements in gait asymmetry that last for an extended length of time (>3 months).

The process of learned adaption can perhaps be considered analogous to studies carried out involving throwing accuracy, with and without prism glasses; functioning to distort vision. Martin et al., (1996) showed that with initial distortion of vision, participants missed the target to one side by a large distance. They were able to correct this after a number of throws reducing the missed distance until they were eventually able to hit the target. When the glasses were removed they then missed the target in the opposing direction. The results show a similar trend of adaption though error feedback as that displayed in this study. After long-term repetition of the experiment, the subjects were able to hit the target under both conditions at the first time of asking. These types of studies differ fundamentally in that adaption to SB does not directly involve visual feedback but the similarities with regard to feedback and motor adaption are evident.

Split belt treadmill walking has been shown to create changes in an individual's gait that persist when returned to normal walking. If this is to be used as a rehabilitant tool this must be used with caution as if the "wrong" leg is placed on the fast or slow belt it may exaggerate the asymmetry of a subject's gait (Reisman et al., 2013). In long-term use of SB treadmill training in stroke patients it was found that 5/13 were deemed to be non-responders (Reisman et al., 2013). This non-response may be due to damage of the essential pathways of neuromuscular control following stroke, or may be due them being unable to adopt a specific strategy to accommodate SB walking. It was shown in this study that subjects adopt specific strategies such as the heel hang to prolong the swing phase of the fast leg. As subjects were noted to adopt varying strategies in accommodation of split belt, perhaps it would be advisable to test, as in this study, the reaction of a subject to short periods of SB walking. The information from a short period of SB exposure can then be analysed to target the use of SB training for long term benefit.

If a certain strategy could be described as the ideal response then the CAREN system may provide an excellent opportunity to re-train gait. The CAREN system has the ability to offer visual feedback to the subject on specific gait parameters; subjects are then able to adapt these parameters to reach a target level (van der Noort et al., 2015). This could be used to encourage the adoption of specific strategies to improve symmetry in gait. The use of this technology may provide further improvements in rehabilitation as subjects could be given real time feedback on asymmetry and thus provide a target or goal, a key component of rehabilitation.

Malone and Bastian (2010) investigated the effect of accommodation in SB walking under conscious thought and with cognitive distractions. It was found that when healthy subjects were distracted, the learning towards accommodation was prolonged but the after effect is also more pronounced, thus possibly providing improved long term adaption for rehabilitation. The virtual reality environment may allow for exploration of further stroke rehabilitation methods (Laver et al., 2012). Learning is a complex assimilation of the senses and so visual flow must also be important. Plotnik et al. (2015) showed that virtual reality visual flow improves the speed of adaption towards a steady walking state in self-paced treadmill mode. Indeed this virtual reality visual flow may account for the slightly more rapid adaption time shown by these results when compared to previous studies using conventional treadmill set-ups (Reisman et al., 2005; Prokop et al., 1995; Dietz et al., 1994, Huynh et al,. 2014). Community ambulation is a complex environment, much more so than a controlled laboratory, with a wide range of distractions and irregularities that make mobility within the community more difficult for those with disability (Shumway-Cook et al., 2002). Visual flow is an important factor in this and should be considered an important component of rehabilitation improving both the adaption process and preparation for community ambulation.

The study had limitations in the sample population, there was a small sample size (n=10) and all subject were young (23.2 ± 0.7), healthy adults. Thus the implications for stroke patient rehabilitation drawn from this research are limited. Similar studies, however, have shown that stroke patients can adapt to SB walking to a comparable extent as healthy subjects (Reisman et al., 2013). Therefore the use of healthy subjects as a model to establish basic principles can be justified to test protocols.

Although sample size is small, there was a rigorous experimental protocol developed that was strictly adhered to. The time of exposure to SB was limited; 15 minutes overall in 10 and 5 minute blocks. This level of exposure is insufficient to result in long term changes in gait as shown by return to baseline after only 2 minutes of return to normal walking. Further studies should investigate the effects of prolonged sessions (>10 mins) of SB walking.

The study did not use force plate analysis, this would have been of benefit in data processing and accuracy the identification of foot strike in the gait cycle. Instead this was defined by observation in analysis, and although carefully assessed this may be liable to human error.

Similar studies utilised a range of split speeds in ratios up to 4:1 (Reisman et al., 2005). This may be of benefit in creating extreme asymmetry and thus greater after effect, although the resulting instability and deviation from normal gait this causes must be questioned. Subjects in the present study walked at 2:1 speeds and, initially at least, this was considered by subjects to be somewhat challenging. As these are healthy subjects with good balance and stability the use of high speed ratios in stroke patients should therefore be treated with caution. During pilot testing of protocol it was decided that returning to the "slow" speed would be most suitable (half that of preferred comfortable walking speed). This was to reduce the instability experienced by subjects in change of conditions, with a return to higher speeds considered to be more unstable and less comfortable. The study was the first of its kind in the CAREN system and so risk to subject was minimised as far as possible. The slow speed, however, was considered by subjects to be uncomfortably slow and may have altered the adaption process in returning to normal walking. Ultimately baseline asymmetries at both fast and slow speeds showed no difference and so this may not be a critical factor.

While it was not quantified, during the experiment it was noticed that a number of individuals appeared off balance during the late single stance phase of the slow limb during SB conditions. This is likely due to the slow speed of the belt as slow walking is associated with greater energy costs that may lead to greater unbalance (Zamparno et al., 1995). During the experiment subjects suffered a number of trips as a result of SB walking. As SB walking requires the subject to remain in the middle of the treadmill with one foot on either belt, any crossover of feet on belts travelling at different speeds resulted in a trip. The subjects were healthy and so were generally able to walk in a straight line and able to accommodate these perturbations without falling. If this were to be used as a method of stroke rehabilitation this must be considered as there would be reduced ability to maintain position on the treadmill without use of a guard between the belts preventing this. There would also be less ability of a stroke patient to accommodate these trips and may result in falls. The CAREN features a safety harness that would prevent falls to ground but it must be considered.

The washout period, in the form of a game and normal treadmill walking may have been improved by a period of over ground walking. Indeed the effect on over ground walking is a topic that deserves further investigation. As ultimately, if to be used in rehabilitation, improved over ground walking is the end goal then this must be tested experimentally in greater detail. As has been shown there is altered biomechanics in over ground and treadmill gait (Alton et al., 1998; Parvataneni et al., 2009; Watt et al., 2009). When the limb is not constrained to belt speed there may result in altered walking, as in rotational treadmill studies in which subject deviated to one side over ground after walking on a rotational treadmill (Gordon et al., 1995). Over ground post SB adaption was investigated by Huyhn et al. (2014) and showed comparable results to the present study.

The adaption to SB walking exhibited significantly less asymmetry and occurred in a shortened number of steps upon second exposure. This may be as the subjects were informed prior to transition and so it may be that there is some form of pre-activation that is found after second exposure as the subjects know what to expect. Further research may also benefit in investigating EMG signals during this adaption process to identify any trends that may be evident.

Previous literature surrounding the use of SB in rehabilitation revolves around the possibility that asymmetry can be improved by the overcompensation experienced when returned to normal walking (Reisman et al., 2013; Malone & Bastian 2014; Torres-Oviedo et al., 2011). It has not been considered that the use of SB walking can force the use of the paretic leg in stroke patients, where loss of bone mass in the paretic limb and overuse damage in the non-paretic limb may occur (Jorgensen et al., 2000; Patterson et al., 2008). As shown in this study the effect of SB does not persist for a prolonged time, although there was limited exposure to SB walking. Constraint-induced movement therapy (CIMT) is a technique used in upper limb rehabilitation that has been shown to be successful in treatment of stroke (Hakkennes & Keating, 2005) and cerebral palsy (Taub et al., 2004). It forces the use of the affected side by restraining the unaffected side. To extend this theory to SB walking, if the paretic leg was placed at a faster speed then it would be required to do more work, perhaps leading to a strengthening of the paretic limb.

Future research should be directed towards extending the experimental protocol to include post stroke patients with asymmetry in gait. It is likely that they will show comparable results. Identification of accommodation strategies in healthy subjects would allow for a target to which stroke patients can receive feedback on and aim towards, possibly improving the adaption process. If SB walking was found to be acceptable to the patients, long term training could be planned in an attempt to improve gait symmetry.

Conclusions

This study shows it is possible to alter symmetry of a person's gait through exposure to SB walking, this asymmetry is then quickly accommodated in young adults as the motor control system adapts to regain symmetry, adopting specific strategies such as heel hang. After a period of prolonged exposure to SB walking the initial asymmetry is reversed when returned to normal walking conditions. This may have significant implications for stroke patients with an asymmetrical gait as a possible method for improving rehabilitation to regain an independent functional gait. It has also been shown that there is a motor learning process that occurs as there is a faster adaption process, shown by reduced asymmetry, upon second exposure to the SB walking condition. This may lead to further insights into the motor learning process and pathways that are related to the retraining gait. The research is unique in its use of the Motek CAREN system to investigate motion capture and SB walking as a rehabilitation method.

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Appendix I

Participant Information Sheet for: Modification of human gait phasing with a split belt treadmill

Biomedical Engineering
Modification of human gait phasing with a split belt treadmill

Introduction
We are a group of researchers at the University of Strathclyde interested in human movement and rehabilitation of stroke. We have a new facility which allows us to measure how people move in a virtual environment; similar to walking with a computer game on a very large screen. Split belt treadmills involve a belt under each leg that can move at independent speed, the body can adapt to this and may result in improved gait in stroke patients. This particular project is being run by Adam Booth (Post Graduate Student, Supervised by Dr. Andy Kerr (Lecturer and researcher working in the Biomedical Engineering Department).

What is the purpose of this investigation?
The study aims to characterise the modification of human gait in response to split belt treadmill walking. This will have implications in the rehabilitation of stroke patients.

Do you have to take part?
Whether you decide to take part or not is entirely your own decision. If you do decide to take part but later change your mind this is entirely up to you and this decision will not have any consequences for you. For Strathclyde students and staff, participation (or declining to take part) in this study will not affect your standing in the University in any way. Individuals should be able to attend an hour long session between 9-5 on designated testing days (June 8th, 16th, 23rd or 30th).

What will you do in the project?
If you decide to take part in the study we will arrange a time for you to come to the University of Strathclyde.. You will need to wear tight fitting clothes (e.g. lycra shorts and top) and soft-soled shoes (for example trainers). We can provide suitable clothing, but you may feel more comfortable in your own clothes. We will attach small reflective markers to your skin using skin sensitive double sided tape and use special cameras to record your movement while you walk. These cameras track the positions of the markers.

Movement on the treadmill involves wearing a harness which will hold you up in case you experience a trip or fall. While you are moving on the treadmill you will be surrounded by a screen showing an outdoor environment which will change as you move. We will ask you to walk at different speeds, One leg will be assigned to the fast speed (which will be twice the speed of the slow leg). You will perform 15 minutes of split belt treadmill training in total, in 5 and 10 minute blocks.

Why have you been invited to take part?
We are looking for a large group of people who have no known issues with their walking. Individuals will be between 20 and 30 years old, generally active, in good health.

Individuals who have a known condition during the investigation period or are taking medication that may compromise their ability to participate are not suitable for this study. Excluding conditions include pregnancy, current ongoing illness, illness within two weeks of session date, exercise intolerance, musculoskeletal surgery (e.g. hip replacement), neurological deficit, vestibular (balance, including travel sickness) problems, and sensory limitations not corrected with glasses or hearing aids.

What are the potential risks to you in taking part?
There is the risk of a fall when walking on the treadmill. You will be secured in a harness when on the treadmill and you will always be supervised when in the laboratory. Sometimes people suffer from motion sickness when moving in virtual environments. If this happens to you, we will pause or stop the session depending on how you feel. Although the double sided tape used to attach the markers is skin sensitive, it can irritate the skin. To minimise the likelihood of this, the markers are only in place for a short period (about 1 hour).

What happens to the information in the project?
We will use a unique code for each individual who participates in the study so all the information you provide and the results of the test, will be kept anonymous. This anonymous data will be kept indefinitely and will be used for research. Data can be withdrawn up until the point it is anonymised.

What happens next?
There will be the opportunity to ask questions throughout the study. If you are happy to be involved in this project, we will ask you to sign a consent form confirming this. If you don't want to participant, thank you for reading this.

Once this investigation is complete the data will be used as a model for stroke rehabilitation. Nothing will be kept that could link the data with any individual.

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Consent Form

Biomedical Engineering
Modification of human gait phasing with a split belt treadmill

- I confirm that I have read and understood the information sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, up to the point of completion, without having to give a reason and without any consequences. If I exercise my right to withdraw and I don't want my data to be used, any data which have been collected from me will be destroyed.
- I understand that I can withdraw from the study any personal data (i.e. data which identify me personally) at any time.
- I understand that anonymised data (i.e. .data which do not identify me personally) cannot be withdrawn once they have been included in the study.
- I understand that any information recorded in the investigation will remain confidential and no information that identifies me will be made publicly available.
- I consent to being a participant in the project
- I consent to being audio and/or video recorded as part of the project

(PRINT NAME)	
Signature of Participant:	Date:

Appendix II

Subject No.	
Group No.	

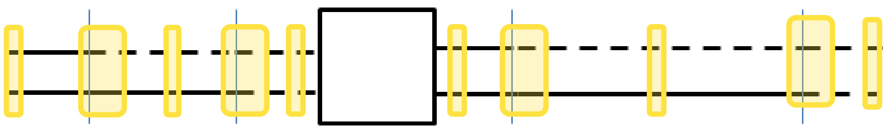
Date:

Time:

Age		Years
Height		cm
Weight		kg
Dominant foot		
Inter-ASIS		mm
	Right	Left
Leg Length (mm)		
Knee Width (mm)		
Ankle Width (mm)		

Preferred Walking Speed (dominant) (Fast)	Half speed (slow)
m/s	m/s

Question	Time Taken	Notes
B–SB1. What stage is normal walking perceived		
SB–B1. What stage is return to normal perceived		
B–SB2. What stage is normal walking perceived		
SB-B2. What stage is return to normal perceived		



Extra Notes: