BIOMECHANICAL STUDY ON LATERALITY DURING MANUAL PUSHING AND LOCOMOTION

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(千葉大学審査学位論文)

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Content

Chapter 1

Introduction to laterality, symmetry and manual pushing

1.1 General Considerations in Locomotion

Locomotion is a common activity in living animal, characterised by movement from one place to another. The methods which animals have utilised to move along the earth's surface seem both diverse and complex, depending on the characteristics of their legs, such as walk, amble, trot, pace, canter, gallop, and hop (Cavagna et al., 1977). In legged animals, the movement is usually conducted with certain distinct patterns of leg movement called as gait and their study is called as gait analysis (Stewart and Golubitsky, 2011). For years, biologists have concentrated on describing different types of locomotion types, and detailed descriptions exist for leg movement patterns and the anatomic features associated with different modes of locomotion (Cavagna et al., 1977). In bipedal humans, most common type of gait are running and walking.

There are various goals of human walking (Knudson, 2007). The first is to move body to a desired place at a desired speed. The second is to achieve the first goal at the least amount of energy expenditure by minimising body movement. The third is to cause the least amount of pain on the painful body parts arranged by brain mechanism to alternate the pressure from the painful body part to the other parts. The fourth is that the foot itself to become a shock absorber for dispersing the force of the body as it lands. The fifth is to make the foot during the end of stance phase to form a rigid lever in order to provide a way to propel the body forward.

As the most energetically demanding daily activities, mechanics of locomotion has attracted substantial research attention (Bertram and Hasaneini, 2013). Walking in humans, involves an alternate transfer between gravitational-potential energy and kinetic energy within each stride. This transfer is greatest at intermediate walking speeds and can account for up to 70% of total energy changes taking place within a stride, leaving only 30% to be supplied by muscles (Cavagna et al., 1977). For decades, a common assumption in comparative biomechanics has been that the walking cooedination should be considered in terms of an inverted pendulum motion during stance that optimises the exchange between kinetic and gravitational potential energy, which meant that most available energy could be saved and re-used in the following stride (Bertram and Hasaneini, 2013). However pendulum model was found to be applicable only at intermediate walking speeds (Cavagna et al., 1977). Cavagna and

Kaneko (1977) reported that above 3 km/h walking speed, internal work, which is the work done on the body segment, is greater than power spent to accelerate and lift the mass of the body at each step. Thus, 3 km/h is the speed suggested as the limitation of internal and external work comparison, as lower than 3 km/h the external work would be greater than internal work.

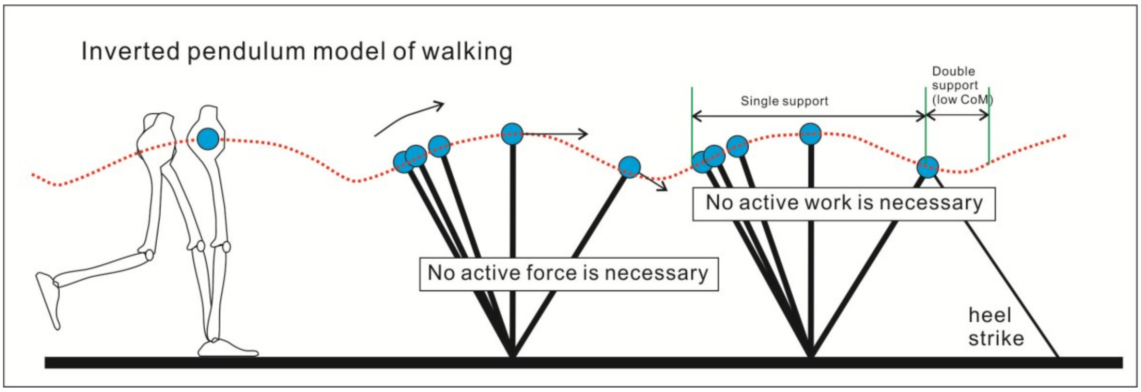


Figure 1-1. Inverted pendulum model during single stance phase of walking (when only one foot contacts the ground). Consecutive single-stance phases are separated by a double stance phase (starting with heel strike), when the CoM (blue circle) is about at its lowest point and one stance leg is replaced by the next stance leg (Kuo et al., 2005).

The energy recovery assumption, however, has not clarify enough the ultimate goal of locomotion study: to understand how adaptations of form and behaviour can influence legged locomotion performance, since the assumption did not take into account the source of energy loss during redirecting body motion from downward to upward at each step (Bertram and Hasaneini, 2013). To compensate this loss, there are two strategies applied: impulsive push-off along the trailing foot right before heel-strike to minimise energy loss, and apply the torque of the hip against the stance leg, using the torso as a fulcrum (Kuo, 2002).

According to Bertram and Hasaneini (2013) there are three main factors that reciprocally influential to determine the most cost-effective movement pattern. The first the energy loss associated with redirecting the centre of mass motion from down to up with each step, which is assumed as the crucial original lost. The second factor is the cost of leg work which is important to control and redirect the centre of mass to help reduce collision loss and replace the loss that remains due to other cause. The third is the cost to swing the limbs. Other factor that is affecting energy loss is step length. The smaller the steps taken, then the downward centre of mass velocity would also be small, thus the collisions which absorb energy are small. However travel with very short steps would necessitates substantial effort to swing the legs very rapidly. In a simple walking model, reducing the step length into half reduces the collisional energy loss in each step by four, and the energy loss in each unit distance in half (Kuo, 2002).

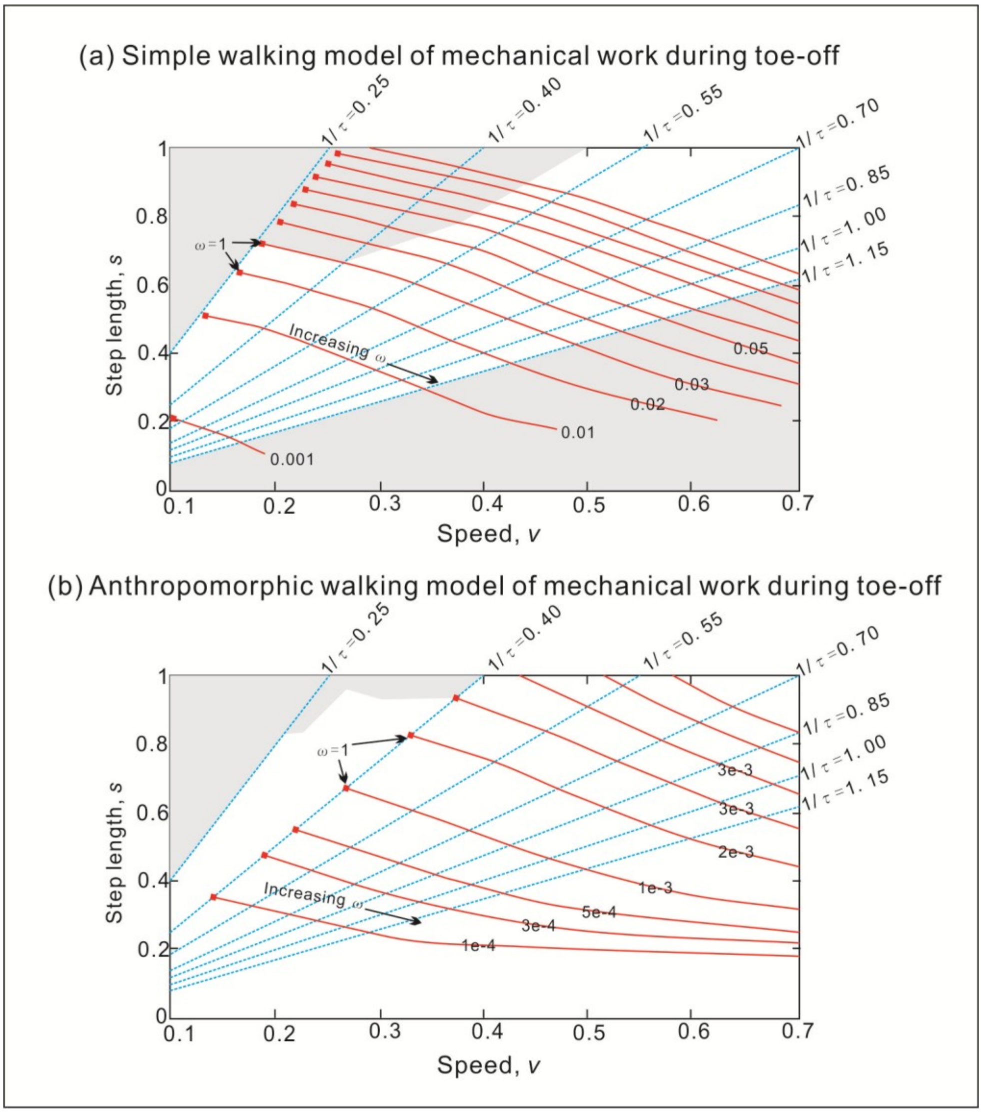


Figure 1-2. Mechanical work generated during toe-off as a function of speed, *v*, and step length, *s*, as showed by (a) the simplest model and (b) the anthropomorphic model. As both speed and step length increase, contour lines of constant (dimensionless) mechanical work per distance (solid red lines, with energy levels, e, as labelled) also increase. Gaits achieved with assumption of the absence of hip spring which is acting on

the swing leg (ω=1, showed by red filled square at the end of the contour lines). Increasing toe-off impulses tend to increase speed and step length linearly, and increasing the swing frequency ω of the leg will affect in the increase of the speed at slightly shorter step lengths. Because the hip spring reduces collision losses, the energy loss during walking is reduced. Step frequency ($1/\tau$, as showed by blue dashed lines) is proportional to ω. The simplest model has regions that are passively unstable (grey areas). The anthropomorphic model, using more realistic body mass properties, generated similar results, however it is actually very difficult to perform very short steps (adapted from Kuo, 2002).

1.2 Physiology of locomotion

Physiological control of movements during manual pushing can be explained in the following description as adapted from description on control of movements by Carlson (1992). When pushing a force plate, the movement of the head and the corresponding movement of the trunk are facilitated by the tectospinal tract, right parietal lobe plays its role in dealing with perception of location of the object being pushed, the information then used by neural circuits in the left parietal association cortex to consider the relative location of the subject's hand and the force plate and send information about the starting and ending coordinated to the left premotor cortex, where the sequential series of muscular contractions required to conduct the manual pushing movement is organized and this sequence is executed through the primary motor cortex and its connections with the spinal cord and subcortical motor systems, the ventral corticospinal tract and the ventromedial pathways (vestibulospinal and reticulospinal system, largely under the influence of bangsal ganglia) adjust posture and control so that subject will not fall, the reticulospinal tract may cause one leg to step forward in order to take bodyweight to generate pushing force, the rubrospinal tract controls the muscle of the upper arm, and the lateral corticospinal tract controls finger and hand movement.

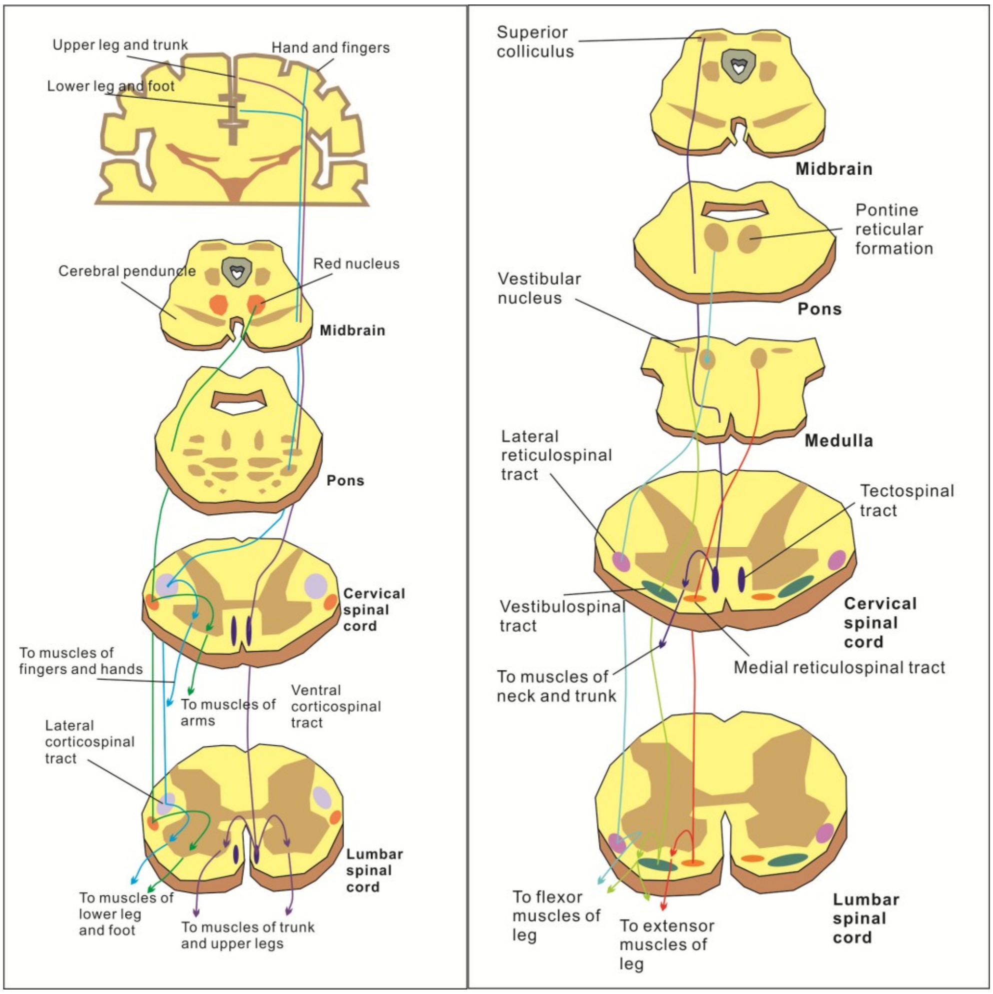


Figure 1-3. The transcortical pathways, cortical pathway, and rubrospinal tract (left), and the ventromedial pathways (right), (adapted from Carlson, 2002).

Human movement is controlled by central nervous system (CNS) using coordination of basic temporal components or activation patterns, shared by different muscles which cooperate to achieve kinematic and kinetic goals, based on both feed forward and feedback signals and produced output in sequential motor task (Lacquaniti, et al., 2012). The timing of the activation patterns is affected by the nature of the task performed and more demanding tasks show higher variability in timing (Lacquaniti, et al., 2012; Oliveira, et al., 2013). Based on both a forward model of limb biomechanics and feedback signals from the periphery, the CNS conducted the strategy to limit the degrees of freedom and issue basic activation patterns which take into consideration the dynamic condition of the whole limbs (Lacquaniti, et al., 2012).

Animal locomotion is controlled, in part, by a central pattern generator (CPG), which is an intraspinal network of neurons capable of generating rhythmic movements of limbs (Golubitsky et al., 1999). However, during evolution, skilled hand movements resulted in the increased influence of a direct cortical-motoneuronal system in parallel with more specialized hand function associated with laterality might have replaced phylogenetically older systems that organized locomotor movements in quadrupeds (Dietz, 2002). While CPG model has been assumed to be symmetry (Golubitsky, et al., 1999), direct cortical-motoneuronal connections especially to hand muscles are thought to determine the degree of laterality in humans and non-human primates (Dietz, 2002).

1.3 Symmetry phenomenon in gait

Symmetry is a phenomenon which is very common to be observed in the universe. In everyday use, the word symmetry and pattern are used interchangeably, to describe the property of more or less identical units' arrangement (Stewart and Golubitsky, 2011). Hermann Weyl described symmetry with two meanings, one to describe a well-proportioned, well-balanced condition, and the other to describe the sort of concordance of several parts by which they integrate into a whole (Stewart and Golubitsky, 2011). In the study of symmetry, there is a paradoxical phenomenon called as symmetry breaking where the symmetric system starts to behave less symmetrically, when the symmetry of the resulting state of the system is a subgroup of the symmetry group of the whole system (Stewart and Golubitsky, 2011). Symmetry breaking is a change in the symmetry group, from the larger one to a smaller one, from the whole to a part.

Symmetry is a transformation instead of a thing, and the main kind of transformation that people thinking about is probably rigid motions of space. The example of spatial motion is translation which is explained as all the points in an object move at the same distance. For example, if we slide a ruler to the left five centimetres, actually all the points in the ruler also move five centimetres to the left. However, symmetries can occur in time as well as in space. Time symmetries are very obvious in a periodic motion which involves changes in time rather than space. The example of time symmetries phenomena are earth movement around the sun in a year and animal locomotion. Any system whose behaviour depends upon time-known as dynamical system-may possess temporal symmetries as well as spatial ones. Even static thing may have temporal symmetries too if we put it into a certain time scale, such as when geologist put a static rock into a study of its changing behaviour on a human timescale.

The patterns of leg-movement are called as gait and the study is called as gait analysis (Stewart and Golubitsky, 2011). In bipedal gait, symmetrical movement during walking of both legs is impossible, since if a person doing so he will fall over. During walking, both legs move half a period out of phase which shows symmetry breaking. In quadrupeds two types of symmetry are observable, namely spatial symmetry which shows the interchanging of fore and hind legs and spatio-temporal symmetry which shows the interchanging of left and right legs with a half-period phase shift, whereas in bipeds, only the latter is observable (Golubitsky et al., 1999). In measuring gait symmetry, two important aspects are commonly used, namely gait cycle which represents the time duration between foot-strike of the same leg and stance phase which represents the time when the foot touches the ground. Stance phase showed as a fraction of gait cycle is called as duty factor (Stewart and Golubitsky, 2011).

1.4 Laterality

The origin of laterality

One example of symmetry breaking phenomenon is laterality. Laterality refers to asymmetrical preferential use of limbs and sensory: handedness, footedness, eyedness, and earedness (Schneiders, et al. 2010). Handedness is found not only in human, but also found in other nonhuman primates (Braccini, et al. 2010; Leca, et al. 2010; Hopkins, et al. 2011; Duarte,et al. 2012) and even in bird species (Hopkins, et al. 2011; Brown and Magat, 2011). Despite handedness as one aspect of laterality has been studied for years, the emergence of laterality during human evolution or during a human lifetime is still less defined, thus attracting the attention of many researchers. Right-handed dominance in the population exists in human population regardless of culture, while culture only affect slight variance in percentage from 74 to 96% among the population were reported to be right-handers (Uomini, 2009). However such consistent right-handedness has not been observed in non-human primates population. Thus human laterality is special because of consistent right-laterality dominance across species level, based on data in various populations separated in geographical area and time, which is not found in other species.

In non-human primates, handedness data are very dependent on contextual observation, such as left-handedness dominance was observed among males chimpanzees in an observation in Tanzania, while the same observation found that females population were more right-handed (Corp and Byrne, 2004), and captive orang-utans (Hopkins et al., 2003), and left-hand preference in order to reach objects such as foods in lemurs, rhesus monkeys, and Japanese macaques. In non-human primates, degree of handedness is also influenced of the complexity of task performed (Uomini, 2009). Based on finding where left-handed dominant was found in orangutans whereas on chimpanzees, gorillas, and bonobos right-handed dominated the population, Hopkins, et al (2011) proposed that the antecedents of right-handedness in human developed in African apes, after they split from the common ancestors with orangutans, and then after the split with chimpanzees, cultural influences resulted in greater proportion of right-handers in human (Uomini, 2009; Hopkins, et al. 2011). However, basically research on non-human primates showed inconsistent patterns across different populations and environments (Uomini, 2009). The most robust evidence of ancient handedness was found in Neanderthal population (Cashmore. et al, 2008; Uomini, 2009) where fossils confirms a modern pattern of the left brain dominance, presumably signally linguistic competence, and asymmetrical arm movements with the traces preserved on the teeth (Volpato, et al. 2012). Studies on mice (Collins, 1968) and twin children (Reiss, et al. 1999; Ooki, 2006; Medland, et al. 2009) found no evidences of genetic influences on handedness.

In general, despite extensive research on laterality, the timing of laterality emergence in the evolution phases remains unclear (Cashmore, et al. 2008) and various methods such as comparison from asymmetry fossils or other great apes in existence have been producing inconsistent results (Marchant and McGrew, 1991; Cashmore, et al. 2008).

Laterality, bipedalism and gait

There are three main theories suggested the origins of bipedalism in humans: Wheeler`s thermal radiation hypothesis, Lovejoy`s provisioning hypothesis, and Hunt`s small-object postural hypothesis. Some other studies suspected that probably there was more diversity in the earlier phases of bipedalism than previously assumed (Hutchinson and Gatesy, 2001; Harcourt-Smith and Aiello, 2004; Friedman, 2006).

Study on chimpanzees found that a bipedal stance without hand support will evoke hand preferences, suggested that the combination of bipedalism and tool use may have helped in developing lateralization in human (Braccini, et al. 2012). Study on Sichuan snub-nosed monkeys also found that foot preference was significantly stronger in bipedal action than in quadrupedal action, suggesting that bipedal posture could be a crucial factor in influencing footedness and handedness (Zhao, et al. 2008).

Laterality, tool adaptation, and manual material handling

The needs to carry food, tools or infants have been hypothesized as a booster for bipedalism which was proposed by Owen Lovejoy in 1980 (Duarte, et al. 2012). Duarte et al (2012) based on research on capuchin monkeys during load-carrying found bent-hip and bent-knee gait similar to other primates walking bipedally, while different patterns were observed during unloaded gait, supporting the opinion that load-carrying is strongly related to bipedalism. Japanese macaques also showed that hand specialization and handedness were stronger in the in the coordinated bimanual stone handling patterns than the unimanual patterns (Leca et al. 2010).

Different from other nonhuman primates, cerebral cortex of modern humans is characterized by certain functional asymmetries, which is an evolutionary consequence of the requirement for asymmetric employment of the forelimbs in the making and using of tools during evolution coupled with linguistic mechanisms (Frost, 1980), supported by earlier research on influence of tool-use on brain development (Beck, 1974).

Current laterality issues

About 85% of human population are right-handed with proportion probably varies according to cultural differences between 74 to 96% (Uomini, 2006). Other suggested around 90% of humans are right handed, around 80% are right footed and around 70% are right eyed (Carey, et al. 2001). Ear preference was found to be more related to footedness than handedness, and probably lateralization is more related to asymmetrical control of whole-body sequenced movement (Chapman, et al. 1987; Peters, et al. 1988; Elias, et al, 1998; Schneiders, et al. 2010). Since cultural limitation is not imposed on footedness, it is probably a more natural predictor of laterality than handedness

(Chapman, et al. 1987; Golomer and Mbongo, 2004; Schneiders, et al. 2010). A study showed stronger relationship between language laterality and footedness than handedness, therefore an equation of handedness and language laterality should also include footedness (Watson, et al. 1998).

Some studies have reported crossed symmetry phenomena where the long bone dimensions of the upper limb showed greater right-biases accompanied by left-biases of the lower limb with smaller lateral differences (Auerbach and Ruff, 2006). The accepted consensus is that adult right-handers have longer (Peters, 1988) and heavier left leg (Chibber and Singh, 1970), while Hatta, et al. (2005) found that stronger left-leg muscle was different according to age. Based on population studies, right foot tends to be the preferred foot of right-handers, however the situation remains unclear for left-handers (Chibber and Singh, 1970; Chapman, et al. 1987; Peters, 1988).

Handedness does not mean one hand is dominant over the other, but more as a complementary role differentiation of high and low frequency tasks execution (Uomini, 2009). Dynamic-dominance hypothesis by Sainburg (2002) suggested that tasks required precision in interjoint coordination and trajectory formation is associated with dominant arm, while tasks less dependent on intersegmental dynamics is less dependent on the dominant arm. In dynamic adaptation to velocity dependent force fields, dominant arm relying on feedforward control while non-dominant arm is relying on impedance control (Schabowsky, et al. 2007).

Population studies on handedness reported some drawbacks possessed by left-handers compared to their right-handers counterpart as reported by Coren (1992). Left-handers have shorter life expectancy than right-handers. Other than cultural and environment limitation, left-handers are also associated with reduced survival ability which included history of birth or pregnancy complications, a slower pattern of growth, sleep problems, immune system difficulties, possible neurological damage, higher susceptibility to a number of diseases, and certain psychopathological conditions such as schizophrenia. Left-handers were also found to have greater risk of accident than right-handers. Left-handers were 20% more likely to have an accidental injury when engaged in sports, 25% more likely to have such injury when at work, 49% more likely to have an accidental injury when at home, and 51% more likely to have accident-related injury when using a tool, machine, or other implement. Left-handers were also reported to be six times more likely to die from causes initiated by accident-related injury (Halpern and Coren, 1991).

In footedness as a bilateral context, the dominant leg is described as the leg mobilized to act upon an object while the non-dominant leg used for postural stabilizing support (Chapman, et al. 1987; Peters, 1988; Hart and Gabbard, 1997; Sadeghi, et al. 2000). Left-footers were found to control their unipedal posture differently from right-footers and footedness influences postural control which result in differences in balance perception and action (Golomer and Mbongo, 2004). However, Hart and Gabbard (1996, 1997) suggested that lower limb choice for postural stabilization in bipedal context may be independent of that in unipedal, where dominant leg is used for more demanding task.

A study on footedness during walking using footprint method found that in the same sex group, footedness does not affect gait parameters such as bilateral step length, stride length, step width and gait angle (Zverev, 2006). In study of gait initiation, Hesse et al. (1997) found that in normal subjects gait initiation is highly symmetrical. This suggestion was negated by research review from Sadeghi, et al. (2000) who formulated a consensus that asymmetrical lower limbs behavior is in existence during gait as a reflection of natural functional differences in propulsion and control, where dominant leg is used for propulsion while medio-lateral balance is controlled by non-dominant leg. This was supported by Dessery et al (2011) who observed asymmetrical frontal body motion influenced by footedness in gait initiation.

The general question in footedness study is whether it is necessary to specialize foot use just like the hand, since the activities of the feet compared to the hands are less complex (Peters, 1988). The study of footedness in general is important for efficient rehabilitation in diseases such as stroke, for understanding neurodevelopmental processes, gait training, and the design of human-machine interfaces (Peters, 1988; Polcyn, et al. 1998; Schneiders, et al. 2010).

Humans possessed a greater difference of functional and structural characteristics between arms and legs compared to other primates, and since footedness measurement is contaminated by postural control, the status of leg and foot control differs from that of arm and hand control (Peters, 1990). Foot tapping test is attractive because it probably the only test where hand and foot performance can be compared directly

(Peters, 1990). While tests for footedness are relatively fewer than that for handedness, soccer is considered useful for footedness study (Carey, et al. 2001; Texeira, et al. 2003).

Laterality issues and physiological anthropology study

In physiological anthropology, the understanding of information on basic human physiological characteristics is very useful in applied sciences and manufacturing industries (Sato, 2005). The development of physiological anthropology along with the advance of measurement methods have proven the importance of this field in explaining phenomena in various activities of human daily life. Furthermore, greater information exchange between researchers around the world has brought better understanding of the conceptual framework of physiological anthropology, as a collection of various studies separated by time and space, into a holistic approach and understanding of the subject (Sato, 2005).

In physiological anthropology, there is a mindset called as individual thinking, which assume that every person has his own uniqueness and special characteristics thus making the idea of ideal man irrelevant (Sato, 2005), in line with Darwin's evolution theory which stated that no two individuals are exactly alike. In living organism, a long evolutionary process has brought varying functional, morphological, and behavioural characteristics due to differentiation of evolution process into various directions, therefore it is necessary to grasp the various physiological functions which exist at the individual's level (Iwanaga, 2005). The measurement in physiological anthropology is based on individual measurements followed statistical treatment of individual data in a group. The measurement is not limited into general measurement of only an individual, but also the smaller components of body organ that compose the individual organism. In our study, laterality is investigated based on physiological anthropology principle by analysing in each subject followed by statistical analyses. While most previous research has studied laterality, especially handedness based on population observations, probably due to greater uniformity among humans across different cultures separated by time and space, compared to other animals, in our study, we consider individual variations carefully, and assume that even in laterality study, such variation exists. The study involving both handedness and footedness like in our study is likely will show this individuals variation to a greater degree because individuals would likely have different degree of handedness and footedness.

Figure 1-4. Research on laterality was initiated by handedness studies from behavioural study perspective, because role difference between left and right hand is very obvious, and can be measured easily based on population study. Behavioural studies which dominate handedness study further began to be implemented in other aspect of laterality such as footedness, eyedness, and earedness. Asymmetry during gait, on the other hand, was not started by behavioural study, but from biomechanics perspective by the introduction of symmetry index by Robinson, et al. (1987) which then, became widely used in measuring gait symmetry. Studies on gait symmetry/asymmetry found that even during bipedal task, asymmetry exists. By considering whole body coordination which is one keyword of physiological anthropology (Sato, 2005; Iwanaga, 2005), in activities such as gait and manual pushing, where left and right side of the body interact each other, asymmetry exists and difference due to laterality could be observed.

Various sorts of coordinative mechanisms act within the bodies of individuals, and to understand how the coordination is performed within individuals is one of main concerns in physiological anthropology studies (Sato, 2005). The example of the mechanism of coordination can be observed in the antagonist and agonist muscles during moving certain parts of an individual body. The whole body coordination, thus, becomes one of the keywords of physiological anthropology (Sato, 2005). This whole body coordination mechanism has been assumed to be necessary to be uncovered in order to better understand differences between individuals. Understanding a phenomenon in a holistic perception is necessary to avoid misunderstanding on various physiological phenomena as independent phenomena separated each other (Iwanaga, 2005), which in reality they are coordinated each other. In handedness studies, it is common to find experimental procedure which separate left and right hand, each was tested by performing the same task to examine degree of handedness. While dominant hand is easy to observe in daily life, footedness is less obvious since compared to the hands, the activities of the feet are less complex (Peters, 1988). Putting whole body coordination as analytical base, in activities such as gait or manual pushing, it seems impossible to isolate hands and feet from coordinating each other, therefore study solely based on either handedness or footedness seems unlikely represents the real situation. Combination of handedness and footedness in many studies have been neglected because most of them ignoring whole body coordination principle. While right-handers and right-footers were found to be the majority of the population, the strength level of laterality are varied and if we put combination of handedness and footedness into account, the variation probably higher. For example, a strong right-hander is probably also a weak right-footer or even a mixed-footer. And since left-handers were reported to be varied in their footedness, their variability is probably even higher. Hence, investigating laterality in whole body coordination framework would likely produce new information which so far has been less explored.

Whole body coordination in physiological mechanisms is accompanied by latent physiological function (Iwanaga, 2005). Human naturally possesses certain level of tolerance to environmental condition, however researches in physiological anthropology found that human capability to adapt to environmental change revealed that human also has potential reserves that would be activated in case hazardous environmental condition push the limit of this tolerance. This field of research is called functional potentiality, which is intended to explore large unidentified human potentialities, with assumption that adaptability should not be separated from human's potentialities. While laterality factor, such as handedness has been associated with adaptability to environment, from physiological anthropology viewpoint, so far has been less explored. Physiological anthropology studies aim to explain behaviour of a living individual in a broad understanding, such as behaviour in its relation with survival and reproduction, and behaviour that is beneficial and less beneficial for the organism in the framework of evolution and adaptability (Iwanaga, 2005). Based on population studies, left-handers were reported to possess various disadvantages compared to their right-handers counterpart, which greatly affect their survivability (Coren, 1992). However physiological anthropology studies to investigate such problems are still very limited. And better understanding on functional potentiality of laterality based on well-defined advantageous and disadvantageous of both left- and right-sided laterality will be a great help to improve human adaptability and quality of life.

1.5 Manual pushing from the viewpoint of physiological anthropology

There are abundant studies on manual pushing with various results with strong focus on industrial ergonomics to reduce the risk of musculoskeletal pain and increase working productivity. The initial studies on pushing were based on static horizontal force exertion in the sagittal plane (Resnick and Chaffin, 1995), with many studies addressing handle height of pushed cart at hip, elbow, and shoulder height (Resnick and Chaffin, 1995; Jansen, et al. 2002; Hoozemans, et al. 2007), and while upper extremities have been studied extensively (Metter, et al. 1997; Voorbij and Steenbekkers, 2001), followed by growing attention on trunk muscles load (Hoffman, et al. 2007), lower extremities have been less studied.

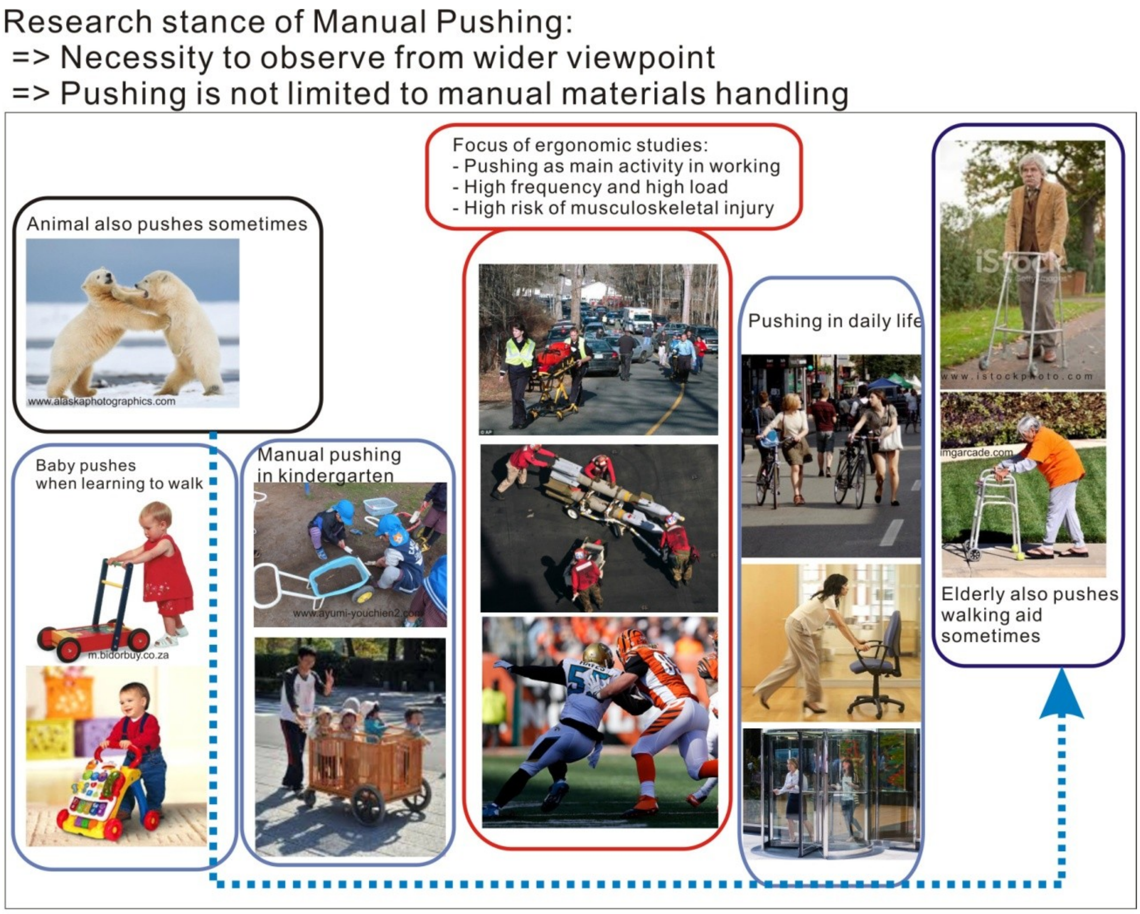


Figure 1-5. Studies on manual pushing have been focused on ergonomic studies, where pushing is involved as main activity, such as in industrial works or sports in order to optimise performance. However, pushing actually exists in many activities in daily life, from toddler who learns to walk until the elderly with walking deficiency. There is also pushing which is less demanding in load being handled but performed frequently such as moving a chair equipped with castors, pushing bicycle, or pushing a rolling door. Putting our insight beyond ergonomic studies will show that there are various aspects of pushing have been less explored.

Manual pushing is very common in daily activities, however people's impression on manual pushing has been largely associated with manual materials handling performed in industrial works therefore it is not uncommon to find some people think that they actually never perform any manual pushing in their daily life. Manual pushing also has been associated with occupation performed by low skilled workers, inefficient in the long run with greater technology development, and sometime is being predicted to be diminishing from modern society. Hence, if we conduct literature review on manual pushing studies, majority of them were focused on ergonomic studies to improve workers performance and productivity as well as to prevent injury. While such studies have been proved to be important in preventing economic loss due to injury and its following consequences such as higher number of absentees, less productivity, as well as higher insurance expenditure, very little if any of such studies tried to analyse deeper the problem in manual pushing based on perspective of human physiological condition.

If we broaden our observation on manual pushing, this activity is actually performed by almost every age in humans, in all civilisations, and also probably has been performed in the long history of human evolution. And most of the manual pushing performed were actually not realized by the performer. For example, not many people will realize that most doors are opened by pushing, most chairs with castors are moved by involving manual pushing, and even in activities like closing desk drawer or putting a book back on the book shelf are actually manual pushing.

In many cases, manual pushing was designed as a replacement for manual lifting, and on the contrary to people expectation, in developed countries manual pushing is probably performed more frequently than in the developing world. In the developed countries, workers' salary is much higher, therefore for manual materials handling operation, less number of workers should be able to handle greater load at higher frequency. This thing is further supported by the availability of supporting industrial base to produce a wide range of tools to perform manual pushing. Most furniture, either in the office, public facilities, or in private residential area are equipped with castors, which ease people to move them from one place to another, accommodated by environmental design which allows the movement with less effort.

Manual pushing involves upper and lower limbs coordination to move the body as well as the load. For people with limited ability to walk in bipedal, pushing is performed to help the lower limbs to support bodyweight. This activity is usually observed in a toddler learning to walk or elderly who needs a walking aid. Considering how common and widespread the practice of manual pushing, it is interesting to find that most manual pushing studies were focused on industrial ergonomics with the goal to increase working productivity and prevent injury while the effects of manual pushing on human physiological condition has been largely neglected. Hazardous nature of manual pushing in industrial work has become common knowledge, especially in musculoskeletal injury induced by this activity. However, the effects of manual pushing other than musculoskeletal complain are still not well investigated.

Research on manual pushing as a part of daily activities in human history, from the point of view of its influence on human physiological condition is still very limited. Better understanding of manual pushing will lead more possibility of its implementation beyond manual materials handling. Based on searching in Google Scholar, no research on the influence of laterality on manual pushing is in existence. In manual pushing, body coordination between lower and upper limbs to produce pushing force exertion probably will provide a different view on relation between footedness and handedness. Our literature searching did not find any previous study on symmetry during manual pushing.

Whereas load carrying task and tool-use has been hypothesized as one of the stimuli for bipedalism (Duarte, et al. 2012; Braccini, et al. 2012), the nature of bipedalism on manual materials handling activities such as manual pushing has not been given sufficient attention. Theory of dynamic dominance (Sainburg, 2002) and complementary role differentiation (Uomini, 2009) in handedness, as well as crossed symmetry (Peters, 1988) in relation between handedness and footedness need to be investigated in manual pushing in which coordination between upper and lower limbs is very important.

Different methods on footedness influence on gait resulted in different findings of either symmetrical or asymmetrical gait (Peters, 1988; Hesse, et al. 1997; Sadeghi, et al. 2000; Zverev, 2006; and Dessery, et al. 2011). The manual pushing experiment measuring the influence of laterality probably will provide a different view on this matter, because it measures coordination of force exertion between upper and lower limbs.

Dominant leg is used for propulsion and non-dominant leg is used for controlling medio-lateral balance (Sadeghi, et al. 2000). If this suggestion were also in existence in manual pushing, this propulsion-balance control sequence should be seen not only on the center of pressure on the ground but also centre of pressure of the pushing force. There should be different features of pushing force in accordance to different nature of walking phases and sub-phases.

This study explored the symmetry and laterality in manual pushing and comparison between manual pushing and normal walking. Laterality studies have been focused on handedness as the most common feature representing laterality, both from unimanual and bimanual tasks experiment. While other aspects of laterality such as footedness has begun to receive attention, in general laterality studies based on activities where left and right side of the body is not separated each other, is still very limited.

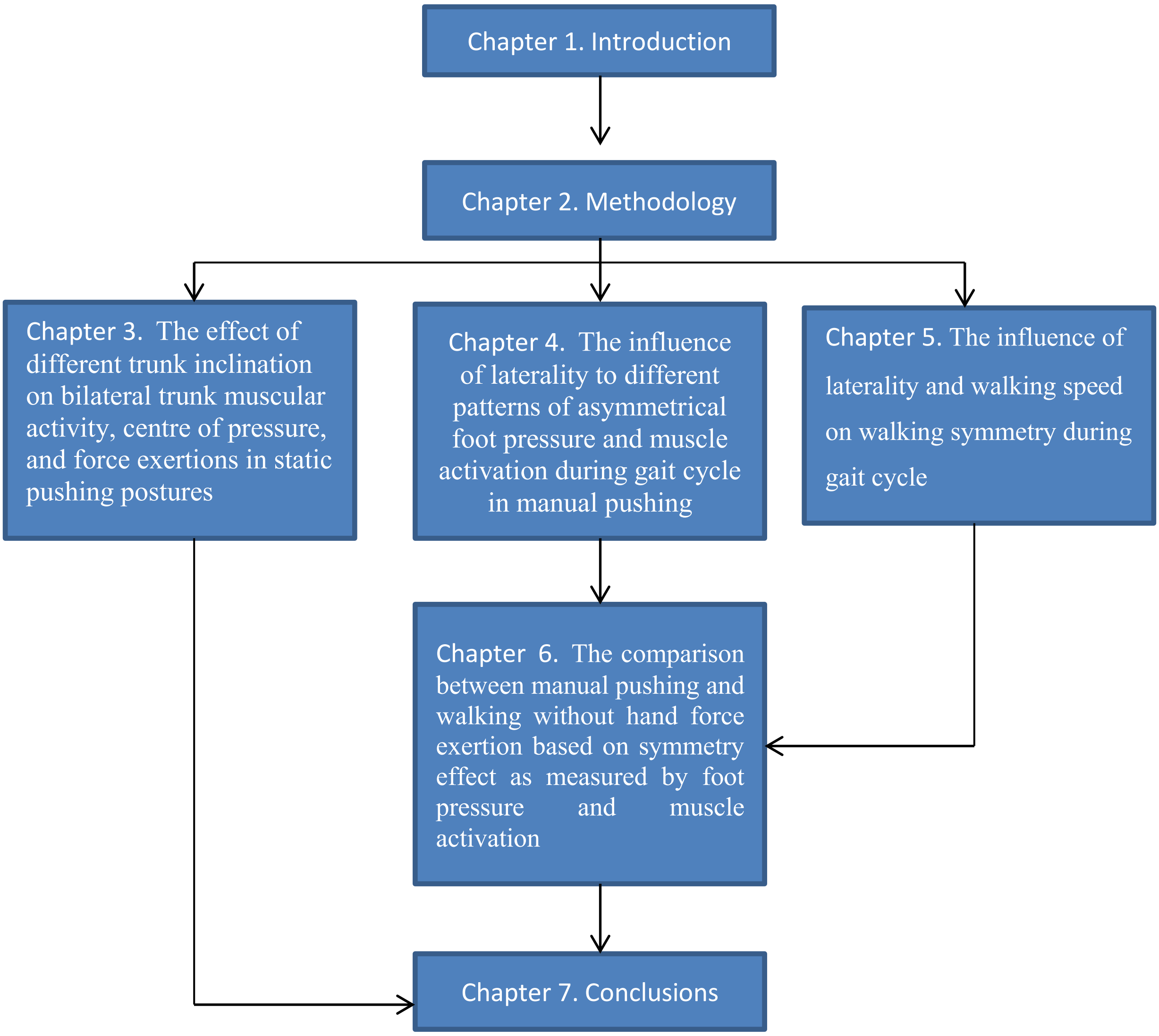
Symmetry and laterality during gait has been studied in the last few decades, especially in relation to walking therapy among patients with walking deficiency or improving performance among athletes. However, to the best of our knowledge, laterality during manual pushing has not been studied. We did not find any explanation of this phenomenon on negligence, however this situation encourage us to explore laterality and symmetry during manual pushing.

1.6 Purpose of the present study

Research on manual pushing as a part of daily activities in human history, from the point of view of its influence on human physiological condition is still very limited. Better understanding of manual pushing will lead more possibility of its implementation beyond manual materials handling. On the other hand, laterality, especially left-handedness has been associated with greater risk to accident or even survivability. However, these studies on risk of left-handedness were mostly based on population observation or psychological study. The analysis on physiological reason why left-handedness is attributed to such hazard is still very limited, and in the study of manual pushing especially almost non-existence. Therefore, this study was intended to investigate symmetry during static manual pushing as well as the relation of footedness and handedness in their influence on manual pushing on a treadmill during gait phases.

1.7 Structure of thesis

The thesis presented here consists of seven chapters. The first chapter includes basic introduction on human locomotion, symmetry concept, laterality, and manual materials handling. The methodology on biomechanical study on locomotion and manual pushing used in this thesis is described in second chapter. The third chapter consists of static manual pushing study, with emphasis on force exertion and trunk muscles activation. The results of this study found asymmetry in manual pushing, which encourage the decision to conduct an experiment on laterality study in manual pushing while walking. The fourth, fifth, and sixth chapters are report from that experiment. The fourth chapter presents the data of manual pushing condition, especially with regard to asymmetry in foot pressure and muscle activation. The fifth chapter presents the data from walking condition. The sixth chapter shows the comparison between manual pushing, walking, and walking holding condition, measured from asymmetry in foot pressure and muscle activation.



The thesis consists of some papers for publication and under preparation manuscript, based on experimental data, which are presented from chapter 3 to chapter 6:

1. Kadek Heri Sanjaya, Soomin Lee, Andar Bagus Sriwarno, Yoshihiro Shimomura, Tetsuo Katsuura (2014) The effect of different trunk inclination on bilateral trunk muscular activity, centre of pressure, and force exertions in static pushing postures. Accepted for publication in Journal of Human Ergology, Vol. 43, No. 1.
2. Kadek Heri Sanjaya, Soomin Lee, Yoshihiro Shimomura, Tetsuo Katsuura (2014) The influence of laterality to different patterns of asymmetrical foot pressure and muscle activation during gait cycle in manual pushing. Accepted for publication in Journal of Human Ergology, Vol. 43, No. 2.

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Chapter 2

Methodology

2.1 Biomechanics

In general term regardless of species, biomechanics is the study of structure and function of organisms by means of the methods of mechanics (Hatze, 1974; Alexander, 2005). However Winter (1990) limited the terminology of biomechanics of human movement as the interdisciplinary science that describes, analyses, and evaluates human movement. At the basic level, the science dedicated to the broad area of human movement is kinesiology, which is a subject combining aspect of psychology, motor learning, physiology as well as biomechanics. Biomechanics is built on the basic knowledge of physics, chemistry, mathematics, physiology, and anatomy. The studies of biomechanics have been initiated centuries ago since the era of Leonardo Da Vinci and Galileo who had interests in applying mechanical principle into biological studies (Winter, 1990).

Biomechanics research includes studies on (a) functioning of muscles, tendons, ligaments, cartilage, and bone, (b) load and overload of specific structures of living systems, and (c) factors influencing performance (Nigg, 1994). Biomechanics as a branch of bioengineering and biomedical engineering can be applied to improve quality of life (Sriwarno, 2008). The physical and biological principles in a wide variety of physical movements from gait to manual materials handling of factory worker to the performance of professional athlete are the same in all cases, the difference are the detail of the specific movement task and the performance of each movement (Winter, 1990).

In ergonomic studies, postural load and musculoskeletal injuries are the primary occupational hazards in industrial work (Sriwarno, 2008). Mittal et al. (1997) reported that manual materials handling such as lifting or carrying load and pushing have been responsible for musculoskeletal injuries such as low back pain, which in the end affects the company financially due to absentees and health insurance. Such injuries may be avoided by studying how the musculoskeletal system adjusts to common work conditions and by developing ergonomic guidelines to assure that manual work conforms more closely to physical limitations of human body and to natural body movements (Sriwarno, 2008).

Knudson (2007) introduced 9 basic principles of biomechanics:

1. Force-motion principle: this principle is based on Newton's laws of motion, with important consideration where an action of a force will change motion.
2. Force-time principle: the duration of time over which force is being applied also has significance in changing motion.
3. Inertia: the feature of an object which resists changes in the object's state of motion, and is also crucial for transferring energy from one body segment to another.
4. Range of motion: the limit of space where body segments move either in linear or angular movement.
5. Balance: an ability to maintain body position with regard to base of support. Stability and mobility of body postures are inversely related, and are possible to be manipulated by several biomechanical factors.
6. Coordination continuum: related to the optimal timing of muscle activation or segmental movement in accordance to the goal of the movement. If high forces generation is necessary, muscle actions and joints rotations are simultaneous, while low-force and high-speed movements are the goal of the movement, muscle and joint actions tend to be more sequential.
7. Segmental interaction: the forces acting in a system of linked rigid bodies consist of various segments interact each other through the links and joints.
8. Optimal projection: for most human movements involving object being propelled through the space there is an optimal range of projection angles for a specific goal. Biomechanical studies show that optimal angles of projection provide the right combination of vertical velocity which determines time of flight, and horizontal velocity which determines range of flight.

2.1.1 Measurement, analysis and assessment

Each quantitative assessment of human movement must be preceded by the measurement and description of phase, and if the diagnosis is significantly required, biomechanical analysis is usually required (Winter, 1990). Previous biomechanical study had the objective to describe a particular movement and any assessment based on visual inspection. Description of the data can take many forms: pen recorder curve, plot of body coordinates, stick diagrams, or simple outcome measures such as gait speed or jump height.

The measurement system produces some data for analysis. Prior to the analysis, it should be confirmed that the data have been free of noise and other unnecessary artefacts. Analysis can be defined as any mathematical operation performed on data in order to present them in different form or to generate a variable that is unable to be measured directly (Winter, 1990). The analysis in EMG involves rectification and filtering. A more complicated biomechanical analysis involves a link segment model, and with kinematic, anthropometric, and kinetic output data can generate significant curves over the course of time which shows large number of kinetic variables such as displacements, joint angles, velocities, as well as accelerations. The biomechanical model is also able to predict net forces and muscle moments that caused the movement by mean of inverse solution technique.

The entire purpose of any assessment is to make a positive decision about a physical movement. Some assessment were based on data analysis, some other were based on a look at the description, especially when the study involves trained observers who can interpret any visual data. The later approach has drawback in that is more speculative and yields a little information regarding the underlying cause of the observed patterns when the trained observers can identify a condition based on the patterns observed but cannot assess why such condition occurred.

2.1.2 Kinematics

An important branch of biomechanics is kinematics, the accurate description or measurement of human motion (Knudson, 2007). Kinematics which is called as the science of motion is a tool to study geometry of movements (Sriwarno, 2008). The term is the English version of A.M. Ampère's *cinématique* (Ampere, 1834), which he created from Greek κίνημα *kinema* "movement, motion". Human movement is measured in a relative position to some reference and is very common expressed as linear (for example: meters) or angular (for example: degrees) units. Angular kinematics is widely adopted in biomechanical measurements because of its adaptability to joint rotations documentation. There are also many other kinematic variables that can be used to record and measure the human motion. Simple kinematic variables are scalars which do not measure direction, while others are vector quantities that also measure the direction of the movement. The rates of change of position measurements based on time derivatives are velocity and acceleration.

Fundamentally, kinematics is based on motion analysis to obtain data of rational objects and coordinate systems. Video camera and body markers have been a very common instrument employed to investigate motion patterns. Basically, individual position of moving objects is identified and acquisitioned using visual motion analysis into an image sequences (Frischholz and Wittenberg, 1997; Sriwarno et al., 2006).

The velocity of a thing is a vector that informs about the direction and magnitude of the change of the position, or in other words, how the position of a point changes with each instant of time. Velocity which is also called speed may be linear (change in position) measured usually in meters per second ($m.s^{-1}$) in international system of units, or angular (change in angle), measured in degrees per second ($deg.s^{-1}$). Average velocity over a time interval represents the the difference of two reference positions being divided by time unit. This average velocity derived from differential of position data is defined in the following equation:

$$Velocity\ (v) = \frac{distance\ moved\ (\Delta d)}{time\ (\Delta t)}$$

Acceleration of a thing is the vector defined by the rate of change of the velocity vector. Just like velocity, acceleration also can be linear (change in linear velocity) with unit of measurement in meters per square second ($m.s^{-2}$) in international system of units, or angular (change in angular velocity), measured degrees per square second ($deg.s^{-2}$). Acceleration is the second derivative of the position vector that defines the trajectory of a particle. It can also be measured directly by an accelerometer which is force transducers designed to measure force associated with a given acceleration (Winter, 1990). The average acceleration of a thing for a certain time period is defined in the following equation:

$$Acceleration\ (a) = \frac{change\ in\ velocity\ (\Delta v)}{time\ (\Delta t)}$$

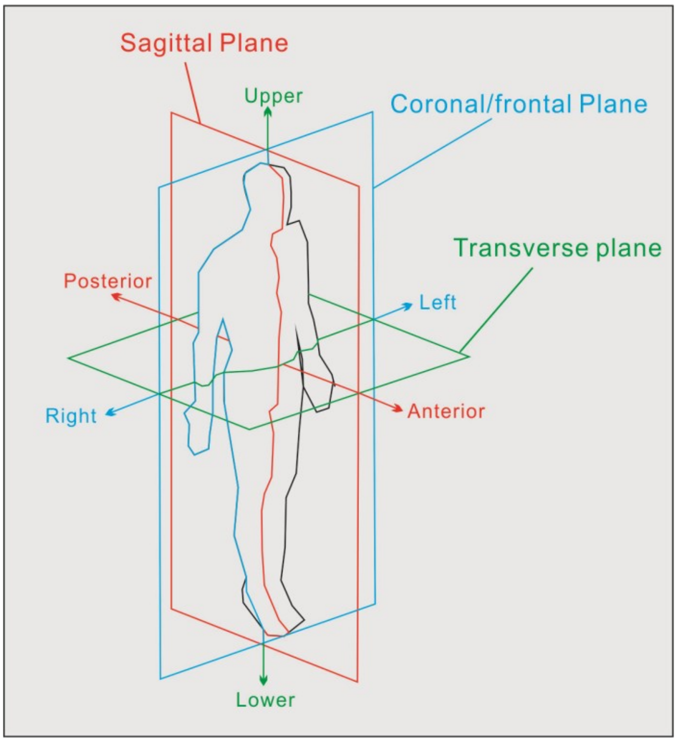

Figure 2-1. Anatomical plane of human body (adapted from www.wikipedia.org).

Anatomical plane of human body:

An anatomical plane is a plane used to cut across the body as a reference to describe the location of structures or the direction of movements. In the assessment of human anatomy, three anatomical planes are used (Wikipedia) as shown in Figure 2-1:

- Sagittal plane, which divides the body into left and right or medio-lateral portions, for example by cutting the axis structures such as the navel or spine.
- Coronal or frontal plane cuts across the body and separate it into dorsal (back/posterior) and ventral (front/anterior) portions.
- Transverse plane, also called as an axial plane or cross-section, cuts the body into cranial (head) or upper and caudal (tail) or lower portions.

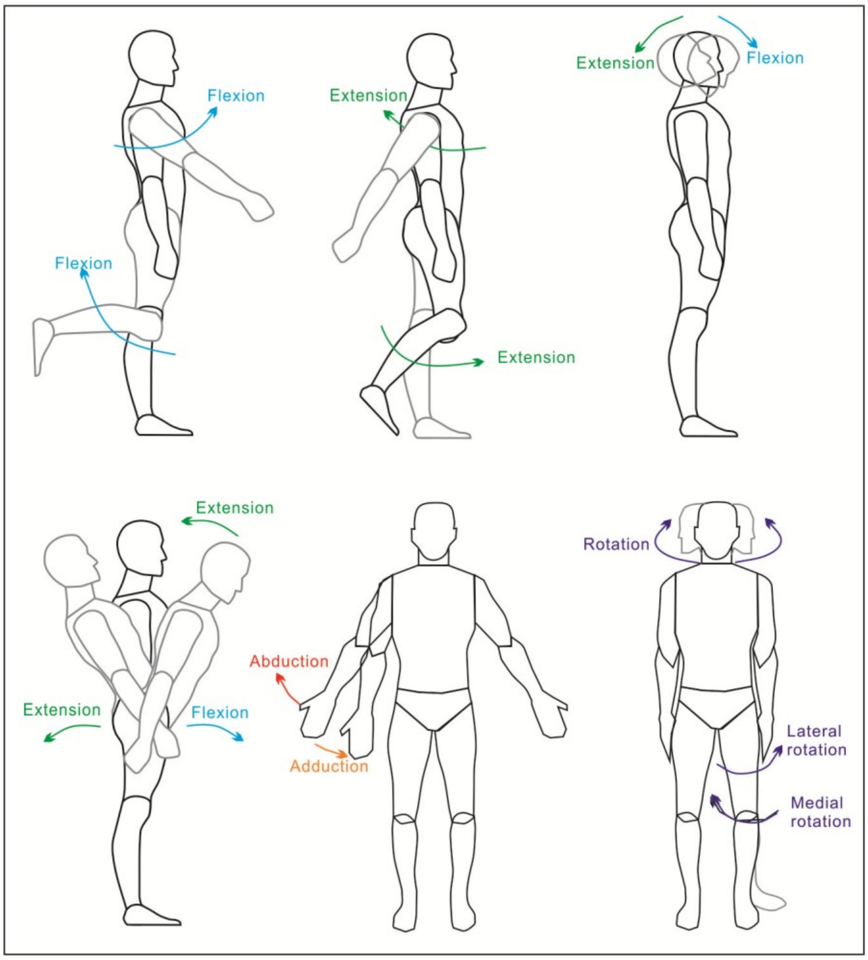

Figure 2-2. Motions in anatomical plane of human body (adapted from www.wikipedia.org)

Motions in anatomical plane (www.wikipedia.org):

- Flexion and extension

  Flexion and extension demonstrate movements that change the angle between parts of the body. If the angle between the two parts is decreasing due to bending movement, this movement is called as flexion. Bending the knee or hip during sitting down, or clenching a hand into a fist, are examples of flexion. In a joint that can move in antero-posterior direction, for example at the neck and trunk, flexion is the forward movement of those parts in the anterior direction (Kendall et al., 2005). Shoulder or hip flexion, on the other hand, refers to movement of the arm or leg anteriorly instead of the mentioned body parts (Cook, 2012).

Extension is on the opposite against flexion, refers to the condition when the angle between the body parts is increasing due to straightening movement. The examples are the movement of neck or trunk backward, or when the knees are extended during standing.

- Abduction and adduction

  Abduction is the movement of a part of the body away from the axis of the body. Especially for fingers and toes, abduction is used to describe the spreading of the digits apart, away from the axis of the hand or foot (Swartz, 2010). In lower limb, when the lower limbs are splayed at the hip, the lower limbs are abducted at the hip (Kendall et al., 2005).

  Adduction refers to a movement of a part of the body closer toward the axis of the body, or in other situation with limb as reference, towards the axis of a limb. Especially for fingers and toes, adduction is used to describe the movement which is bringing the individual finger or toe together, towards the axis of the hand or foot (Swartz, 2010).

- Rotation

  Rotation refers to movement to circle around an axis. Rotation of body parts is divided into internal or external rotation, where internal rotation has direction towards the body axis, while external rotation has direction away from the body axis (Swartz, 2010).

- Foot motions

  - Foot flexion and extension

    Extension or flexion of the foot at the ankle is called dorsiflexion and plantarflexion. When the toes are brought closer to the shin and the angle between the dorsum of the foot and the leg is decreasing, this movement is called as dorsiflexion (Chung, 2005). The most common example of dorsiflexion is observed during walking on the heels-such as during using high heels shoes- when the ankle is being dorsiflexed. Technically extension of the ankle joint refers to dorsiflexion, which could be considered in the contrdictory to the original meaning of extension, as the motion of dorsiflexion reduces the angle between the foot and the shank (Kendall et al., 2005).

When the movement decreases the angle between the sole of the foot amd back of the leg, the movement is called as plantarflexion. Examples of plantarflexion include the movement when pressing a pedal or standing on the tiptoes or calf raises (Kendall et al., 2005).

- Inversion and eversion

  As shown in Figure 2-3, inversion and eversion are the movements that tilt the footsole farther from (eversion) or closer to (inversion) the axis of the body (Swartz, 2010). Eversion is the movement on the frontal plane in which the plantar aspect of the sole of the foot is tilted away from the axis (Chung, 2005). Inversion is the movement of the footsole on the frontal plane where the plantar surface is tilted to face the axis of the body or the medial sagittal plane (Chung, 2005).

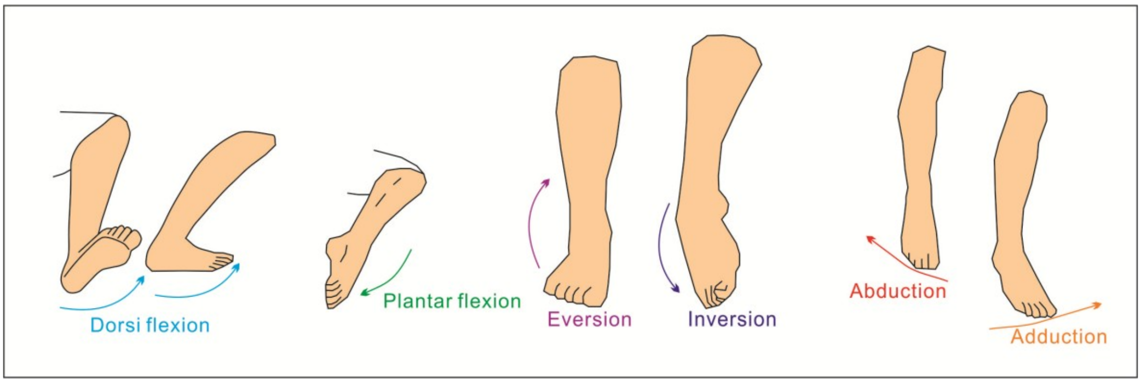

Figure 2-3. Foot motions (adapted from www.courses.vcu.edu)

2.1.3 Kinetics: forces and moments of force

Different from kinematics, kinetics as a discipline studies motion with regard to the sources that cause the motion with measured variables such as force, momentum, and energy (Chapman, 2008). Linear kinetics is the study of the forces that is the causes of linear motion (Knudson, 2007). There are several laws of mechanics that can be applied to a study of the causes of linear motion: Newton's laws, the impulse–momentum relationship, and the work–energy relationship. The most common approach involves Newton's Laws of Motion, called the laws of *Inertia*, *Momentum/Acceleration*, and *Reaction*.

Newton Law of Motion:

First Law: if there are no forces acting upon a particle, it will remain either in static condition or move in a straight line with constant velocity. Newton put the first law of motion to develop frames of reference for the application of the other following laws. The first law of motion assumes the existence of frame of reference called a Newtonian or inertial reference frame, a reference that is relative to which the motion of a particle not subject to forces is a straight line at a constant speed. Newton's first law is often referred to as the law of inertia.

$$\sum F = 0 \Leftrightarrow \frac{dv}{dt} = 0$$

Second Law: Acceleration is produced when a force acts on a particle with certain amount of mass, where the vector sum of the forces is equal to the mass of that particle multiplied by the acceleration vector. The second law states that the net force on a particle is equal to the rate of change of its linear momentum in an inertial reference frame:

$$F = \frac{dp}{dt} = \frac{d(mv)}{dt}$$

The second law can also be stated with regard to a particle's acceleration. Newton's second law is only applicable for constant-mass systems, mass can be taken outside the differentiation operator by the constant factor rule in differentiation. Thus:

$$F = m\frac{dv}{dt} = ma$$

Third Law: When two particles exert force upon another in the same line, the forces act along the line joining the particles and the two force vectors are equal in magnitude and opposite in direction. The third law means that all forces are interactions between different bodies, and thus that there is no such thing as a force that acts on only one body. This law is sometimes referred to as the action-reaction law.

$$F_{action} = -F_{reaction}$$

Inertia is a property of an object by which it has the tendency to resist changes in their state of motion in a straight line (Knudson, 2007). The Inertia Principle suggests that the smaller mass the easier the object to accelerate, while greater mass will increase the stability of the objects and makes it harder to accelerate. Applying the Inertia Principle might also mean using more mass in activities where there is time to overcome the inertia, so that it can be used later in the movement. When two objects are contacting each other, the forces generated by the interaction between the bodies are resolved into right-angle directions: normal reaction and friction. The Impulse–Momentum Relationship says that the change in momentum of an object is equal to the impulse of the resultant forces acting on the object. This is Newton's second law when applied over a time interval. The real-world application of this relationship is the Force–Time Principle.

Energy describes the capacity to do mechanical work; in biomechanical studies, energies include strain, potential, and kinetic energy (Knudson, 2007). The Work–Energy Relationship states that mechanical work is equal to the change in mechanical energy. Mechanical power refers to the rate of work being performed, and can be calculated based on the product of force and velocity. The Segmental Interaction Principle states that energy can be transferred from one segment to the other segments.

Muscle is both the source of mechanical energy generation and major site of energy absorption. A very small fraction of energy is dissipated into heat as a result of joint friction and viscosity in the connective tissues. Thus mechanical energy is continuously flowing into and out of muscles and from segment to segment (Winter, 1990). During walking, there is no external load, and all the energy generation and absorption are required simply to move the body segments themselves. The work done on the body segments is called internal work, and the work done on the load is called external work (Winter, 1990). In manual pushing, the external load is well defined by the pushed object. In walking external work is required to lift body weight into a new height.

Walking is characterized by centre of mass motion similar to that of an inverted pendulum, where during the single support phase of human walking, for example, the stance limb behaves much like a rigid strut allowing kinetic energy to be stored as gravitational potential energy and then returned in a nearly conservative manner (Cavagna et al., 1977). For compatibility with this model, the double support phase is perhaps best regarded as a transition from one inverted pendulum to the next (Donelan et al., 2002). During single support phases, the centre of mass moves along an arc dictated by the stance limb and the centre of mass velocity is approximately perpendicular to the limb. Each transition to a new stance limb requires redirection of the centre of mass velocity from one inverted pendulum arc to the next. Since ground reaction forces are directed approximately along each leg, this redirection of the centre of mass velocity requires negative work by the leading leg beginning at heel contact. To maintain a steady walking speed, positive work is needed to replace the energy lost due to this negative work. The positive work can be performed at any time during a step. But, a simple model of bipedal walking predicts that it is most advantageous for the trailing limb to perform the required positive work at the same time that the leading leg performs the necessary negative work. If the positive work is performed substantially prior to the double support phase, more energy is lost in redirecting the centre of mass velocity. As a consequence, more positive work is required to maintain the same walking speed.

External work which is the work performed by external forces-mainly from the ground-to move the centre of mass through a displacement is more common to be used in locomotion study because of its simplicity since it can be estimated by measuring ground reaction force alone given the knowledge of the speed of the animal (Cavagna, 1975). Internal work measurement, on the contrary, is much more complicated as it requires to mesurements of kinematics and inertial properties of body segments (Donelan et al., 2002). Some methods for measuring internal work are the change of kinetic and potential energy in body segments, centre of mass approach, joint power and work, muscle power and work, and isometric work against gravity (Winter, 1990). Centre of mass approach, based on the potential and kinetic energy of the body's centre of mass is one of the most common method applied, however Winter (1990) suggested that this method has deficiency where it does not account for the energy losses due to the simultaneous generation and absorption of energy at different joints.

In our study, to measure force we used force plate consists of several load cells, which is shown in Figure 2-4 (left), that provide information about total force acts upon the load cell. The combination of several load cells in a force plate also provides information on the movement of centre of pressure (CoP) which has been very common used to measure balance and stability in postural control.

Figure 2-4. Left picture: Load cell (Kyowa Corp., Japan); right picture: pressure sensor (Interlink electronics, USA).

Foot sole of human which is characterised with narrow width and greater length causes the ground contact of various points on the foot sole do not occur simultaneously. In order to measure the ground contact timing we used pressure sensors, as shown in Figure 2-4 (right), attached on five points of the sole of each foot. This provides information not only the timing, but also the magnitude of pressure which is changing along the time course.

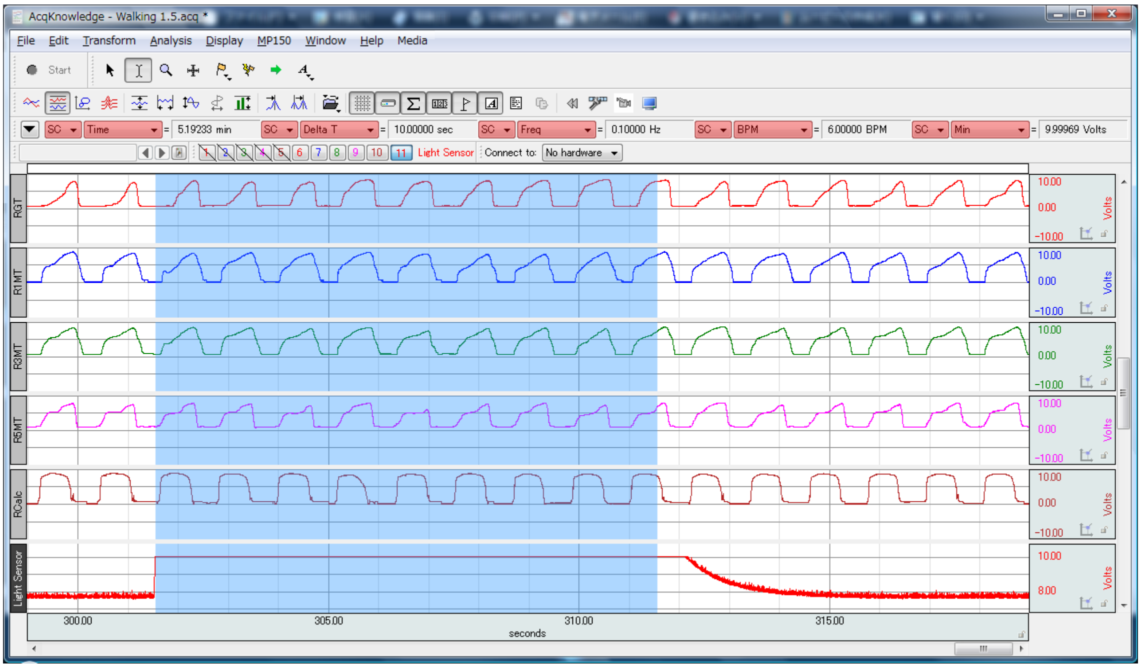

Figure 2-5. Foot pressure sensor measurement was selected in 10s, from the beginning of light sensor recorded the light stimulus to mark the beginning of the trial.

As shown in Figure 2-5, the foot pressure sensors provides signal of pressure bilaterally from big toe, 1st, 3rd, 5th metatarsals, and calcaneus. To measure the timing of recorded trial, we used a light sensor which was working based on light stimulus. The stimulus also marked the onset of the trials for each subject. In the figure, the trial was recorded for around 10s, from which the data processed for further analysis was selected.

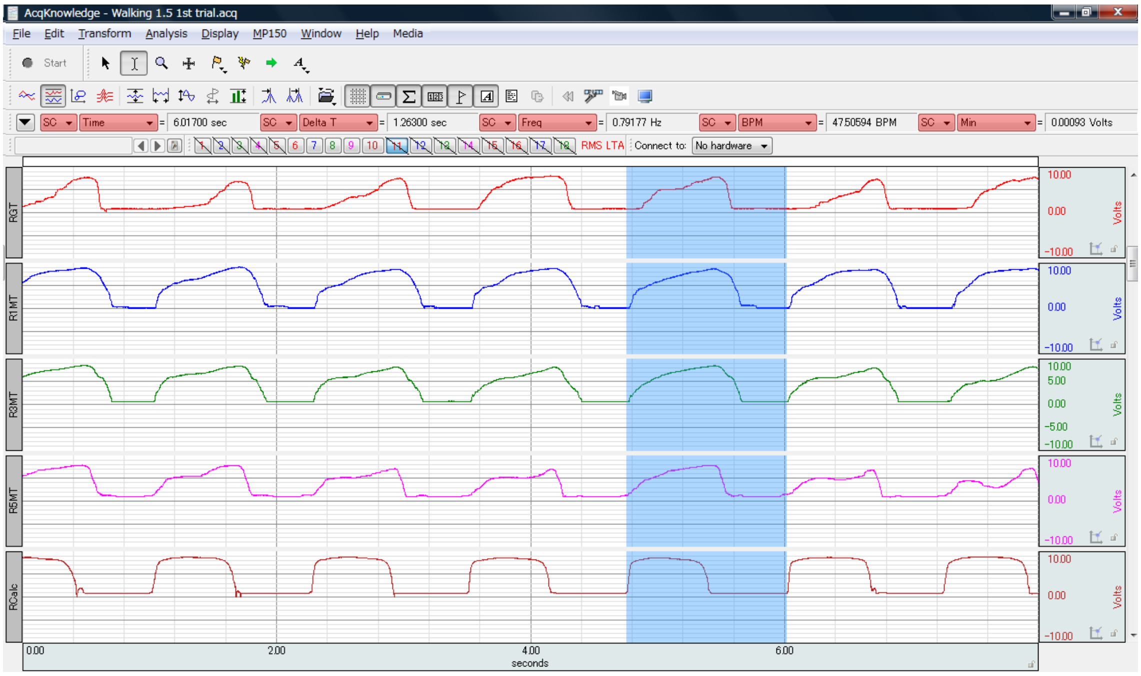

Figure 2-6. Gait cycle was identified by calcaneus pressure sensor, from the initial ground contact to right before the next ground contact.

To ease data analysis for each time unit, in gait studies, most data were normalized into gait cycle which is the initial ground contact to right before the next ground contact of the same foot. As shown in Figure 2-6, in our study we used calcaneus pressure sensor as the reference for gait cycle since most ground contacts were initiated by heel-strike. Gait cycle is divided into stance phase and swing phase. Stance phase is the time when the foot contact the ground and swing phase is the time when the foot is off the ground, also called non weight bearing phase, when the foot swing from one stance phase to another stance phase. In Figure 2-7, stance phase duration is based from the total duration of the five pressure sensors in one foot. The pressure sensors also provide the timing of each sole point contact relative from each other.

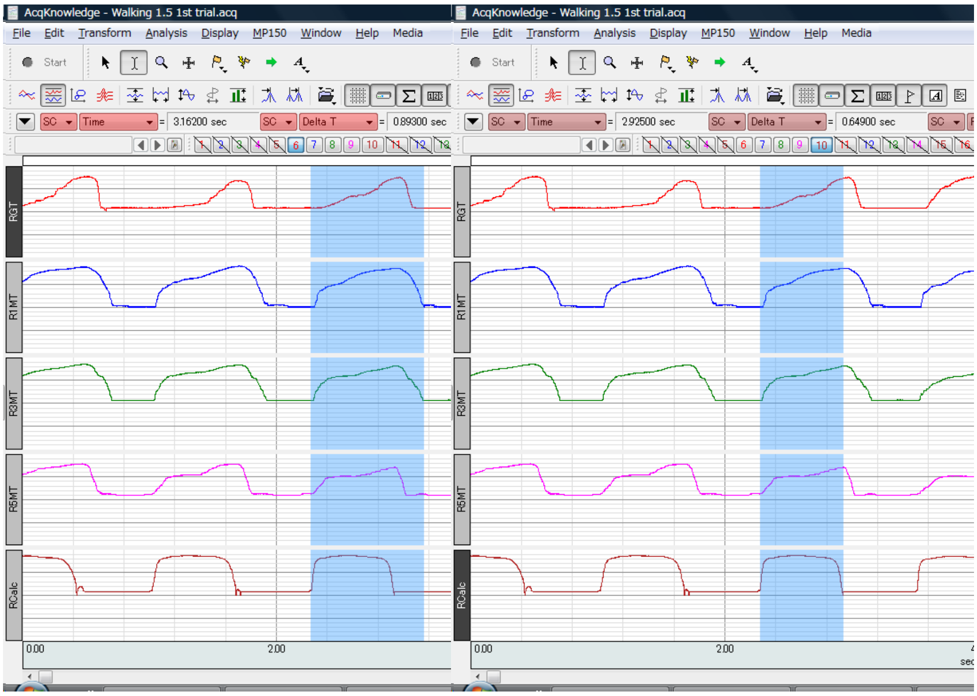

Figure 2-7. Stance phase duration was identified by the total duration of all pressure sensors (left picture), and each of the five points contact duration was based on the duration of the spike of the signal from each pressure sensor.

2.1.4 Anthropometry

As sourced in Wikipedia, anthropometry (from Greek ἄνθρωπος *anthropos*, "man" and μέτρον *metron*, "measure") refers to the major branch of anthropology that studies the physical measurements of the human body to determine differences between individuals. Anthropometry was a fundamental instrument of physical anthropology, because it has been very useful for identification in order to understand human physical variation, in paleoanthropology and in various attempts to correlate physical with racial and psychological traits. While in the past the emphasis of anthropometry was evolutionary and historical, today, anthropometry is very influential in design and architecture to optimize product and artificial environment. Other than the application of anthropometry in basic linear, area, and volume measures, it also has significance in human movement analysis since kinetic and kinematic measures require sufficient information on subjects' anthropometry data as well.

2.2 Electromyography

2.2.1 General consideration

Muscle is a soft tissue found in most animals which is responsible primarily for postural control, locomotion, and movement of internal organ. The word muscle came from Latin *musculus* which means little mouse, since the movement of some muscles were thought to resemble the movement of mice. Mammals have three muscles: skeletal muscle, smooth muscle, and cardiac muscle (Carlson, 1994). The skeletal muscle is responsible for our movement and behaviour, smooth muscle is found in the gastrointestinal system, uterus and small blood vessels, and cardiac muscle is found in the heart (Carlson, 1994). In this study, we only measured skeletal muscle for locomotion and postural control.

When muscles are activated, they generate electrical currents through the tissues, and the voltage generated may be recorded as myoelectrical signal which represents the mechanical contributions of muscles to anatomical movements (Loeb and Gans, 1986). Electromyogram (EMG) shows information regarding the final control of each signal muscle (Winter, 1990). One of the most common type of EMG instrument is surface EMG which has three main applications: timing of muscle activation, amplitude of force produced by the muscle, and as an index of fatigue processes (De Luca, 1997).

Muscle contraction is defined as the condition when the muscle either draw together, gets shortened or undergo increase in tension (Faulkner, 2003). Based on either length changes or force level generated, the muscles have the ability to contract in various types of contractions namely concentric contraction, eccentric contraction, isometric contraction, isotonic contraction, and isokinetic contraction (Faulkner, 2003; Sriwarno, 2008).

- Concentric contraction: the muscle shortens when it contracts. This is the most common contraction in daily life and therefore what most people think about muscle contraction.
- Eccentric contraction: the muscle lengthens when its contracts, such as when decelerating the movement of body part.

- Isometric contraction: the muscle remains in the same length during contraction.
- Isotonic contraction: the muscle tension remains constant during contraction despite of length changes.
- Isokinetic contraction: the muscle contracts in the same speed.

Muscle is a challenging area in the study of biomechanics and neural control, metabolism, and biomechanical characteristics of muscle are subjects of continuing studies (Winter, 1990). The smallest subunit of skeletal muscle is called as motor unit which is a set of muscle fibers that are innervated by a same motor neuron (Sriwarno, 2008). Motor unit consists of a synaptic junction in the ventral root of spinal cord, a motor axon, and a motor end plate in the muscle fibers. The number of muscle fibers in a motor unit is varied between three muscle fibers in small muscle requiring fine control to as many as 2000 muscle fibers in motor unit of larger human skeletal muscle (Feinstein et al., 1955; Sriwarno, 2008). A muscle fiber is about 100 µm in diameter and consists of fibrils about 1 µm in diameter (Winter, 1990).

There are two ways in increasing tension during muscle contraction: increase firing rate of motor unit or recruitment of additional motor unit (Wani and Guha, 1975). Recruitment of motor units during contraction is based on size principle, the smallest is recruited first and the largest motor unit is recruited last (Henneman, 1974). One factor where force of muscle contraction depends on is the number of fiber activated, therefore a large motor unit which consists of a large number of fibers can exert more force than a small motor unit (Sriwarno, 2008). The larger the motor unit, the larger the motor neurons that innervateand the greater the depolarization potentials seen at the end of the motor plate, and the greater mass of the motor unit, the greater the voltages seen at an electrode of EMG (Milner-Brown and Stein, 1975).

There are two types of motor units: the smaller slow-twitch motor units called as tonic units (type I) and the larger fast-twitch motor units called phasic units (type II) (Wormolts and Engel, 1973; Winter, 1990). The type I is metabolically rich in mitochondria, highly capillarized, and mechanically produces twitches with low peak tension and a long time to peak (60-120 ms). Type II is less rich in mitochondria, poorly capillarized, and produces larger peak tension in a shorter time (10-50 ms). Muscles of the upper limbs usually have shorter time to reach a maximum (for example: triceps brachii: 44.5 ms in average, 16-68 ms in range) compared to lower limbs muscles (for example: tibialis anterior: 58.0 ms in average, 38-80 ms in range; soleus: 74.0 ms in average, 52-100 ms in range).

Muscle consists of tubular muscle cells called myofibers, and muscle cell consists of repeating sections of sarcomeres. Sarcomere is the functional unit of muscle contraction (Loeb and Gans, 1986), which appear in dark and light bands visually, and composed of long fibrous proteins that slides past each other when the muscle contracts and relaxes. There are three processes occur in this level influence the force output of the muscle:

- The length of the sarcomere is directly related to the overlap of the thick and thin filaments, thereby determining an upper limit on the number of cross-bridges located within binding range of activated sites. The degree of overlap is commonly shown in length/tension curve, which usually measured during isometric contraction.
- The force generated by a bound cross-bridge changes dynamically if there is relative motion between the thick and thin filaments, which means the length of the fiber is not constant or isometric.
- The kinetic of calcium release, diffusion and resorption accompanying each action potential constitute the limiting factor in the number and time course of activated binding sites.

2.2.2 EMG signal recording

One most common type of EMG instruments is surface EMG which is considered advantageous than indwelling EMG, because of its non-invasive application is easy to use and able to pick up larger motor zone (Komi and Buskirk, 1970). Surface EMG also records longer duration of motor unit action potential because of larger surface area (Basmajian, 1973; Winters, 1990).

Figure 2-8. MP 150 system (left) and TSD 150A surface EMG electrode (right)

In our study we used Ag/AgCl bipolar electrode. A bipolar electrode which usually has two electrical contacts is designed to measure the voltage difference between between two points relative to a common ground electrode (Loeb and Gans, 1986). The common ground electrode should be positioned on a bony prominence. The dipole form of bipolar electrode has two contacts that are similar each other and separated by some distance. The location, spacing and orientation of those contacts are critical in EMG recording. To achieve optimum recording, electrode should be placed between a motor point and the tendon insertion or between two motor points, and along the longitudinal midline of the muscle (De Luca, 2002) or at the point usually called as muscle belly. The placement should be arranged so that the detection surfaces intersect as many muscle fibres as possible. Ideally target muscle identification is by means of electrical stimulation or surface electrical mapping (De Luca, 1997). However in our study, target muscle identification is based on palpation method where the subject is required to take certain posture so that the target muscle contract and experimenter then palpate the muscle to identify muscle location and estimated size (Muscolino, 2009). The palpation method is easy and cheap compared to other methods.

Figure 2-9. Raw EMG data recording at 1000 Hz. The light sensor marked a 10s recorded trial as shown by the selected data shown in blue color.

It is important that the electrode can make a good contact with the skin. Before attaching the surface electrode, the skin should be shaved and cleaned with abrasive paste and alcohol (Mamaghani, 2002; Sriwarno, 2008). The gel coated electrode then being attached by giving slight pressure and fixed by tapes with sufficient mechanical stability since it is critical in signal acquisition (De Luca, 1997).

Major considerations when specifying the EMG amplifier are: gain and dynamic range, input impedance, frequency response, and common-mode rejection ratio (Winter, 1990). The amplitude of surface EMG has maximum amplitude of gain of 5 mV peak to peak (Winter, 1990). High input impedance is necessary to assure constant system gain with varied electrode impedance (Mamaghani, 2002). It is desirable to have input impedance of 1 M Ω or higher and to prepare the skin to reduce the impedance to 1000 Ω or less (Winter, 1990). Nowadays, all analyses and applications of surface EMG are accomplished with algorithms implemented on computers (De Luca, 2001). Since the algorithms requires analog to digital signal conversion, the concept of sampling frequency arises. It is important that the minimum acceptable sampling frequency in order to correctly reproduce the original analog information (De Luca, 2001). The usable energy of the EMG signal is limited between 0 to 500 Hz frequency range with the dominant energy being in the range 50-150 Hz (De Luca, 2002), therefore we used 1000 Hz sampling rate in our study. Since human body is a good conductor and prone to electrical noise, high common-mode signal rejection ratio (CMRR) to reduce electrical noise from 50 or 60 Hz is necessary, preferably above 80 dB (De Luca, 1997; Mamaghani, 2002).

2.2.3 EMG data processing

EMG raw data may not be suitable for further analyses such as analyses for comparison or correlation with other physiological and biomechanical signals therefore it is necessary to change the raw EMG data into another processed form (Winter, 1990). Any interpretation of raw EMG signal with respect to the force production of the muscle, its relative activation, or its fatigue state is difficult and processing EMG data is important and usually performed either in time or frequency domain (Mamaghani, 2002; Sriwarno, 2008).

Rectification

The full-wave rectification generates the absolute value of EMG, usually with positive polarity by eliminating all the negative values (Winter, 1990). While the original raw EMG data has a mean value of zero, the full-wave rectified signal does not cross into below zero, therefore has an average that fluctuates with the strength of muscle contraction. Typically, full wave rectification is preferred since it retains the entire signal which is often used for further signal processing (Mamaghani, 2002; Sriwarno, 2008).

Smoothing

The full-wave rectified signal still contains high-frequency signals which is removed through low-pass filtering, to obtain a so called linear envelope (Winter, 1990; Mamaghani, 2002). It can be described as a moving average because it follows the trend of the EMG and related to contractile features of the muscle since it quite closely resembles the shape of the tension curve.

Integration

The integrated EMG (IEMG) has been associated to muscular force more than any other form of the processed EMG (Bigland and Lippold, 1954; Thorstensson et al., 1988; Mamaghani, 2002; Sriwarno, 2008). The purpose of the integration process is to measure the area under the curve (Winter, 1990). In this process, the rectified waveform is mathematically integrated with respect to time. The integrated signal will always increase steadily as long as any EMG activity is present, therefore integration is usually performed either over a small time period or the integrator reset to zero when the integrated value reaches a specified limit (Mamaghani, 2002; Sriwarno, 2008). The advance in electronics device have made it possible to develop a newer version of IEMG called as average rectified (avr) value of the EMG signal, which like IEMG, provides of the measurement of the area under the signal but do not have specific physical meaning (De Luca, 2002).

Root Mean Square

The root mean square (RMS) value of EMG signal is a set of values in a continuous waveform that are calculated by summing the squared values of the raw EMG signal, determining the mean of the sum, and taking the root mean square of the mean so obtained:

$$RMS = \left(\frac{1}{T} \int_{t}^{t+T} EMG^2 (t) dt \right)^{1/2}$$

(Mamaghani, 2002)

RMS value represents the signal power, thus it has advantage than integrated EMG in a clearer physical meaning and preferred in most applications (De Luca, 2002). The use of RMS is considered new in the field of EMG measurement since it is allowed by the advances made in the electronic devices in the last few decades.

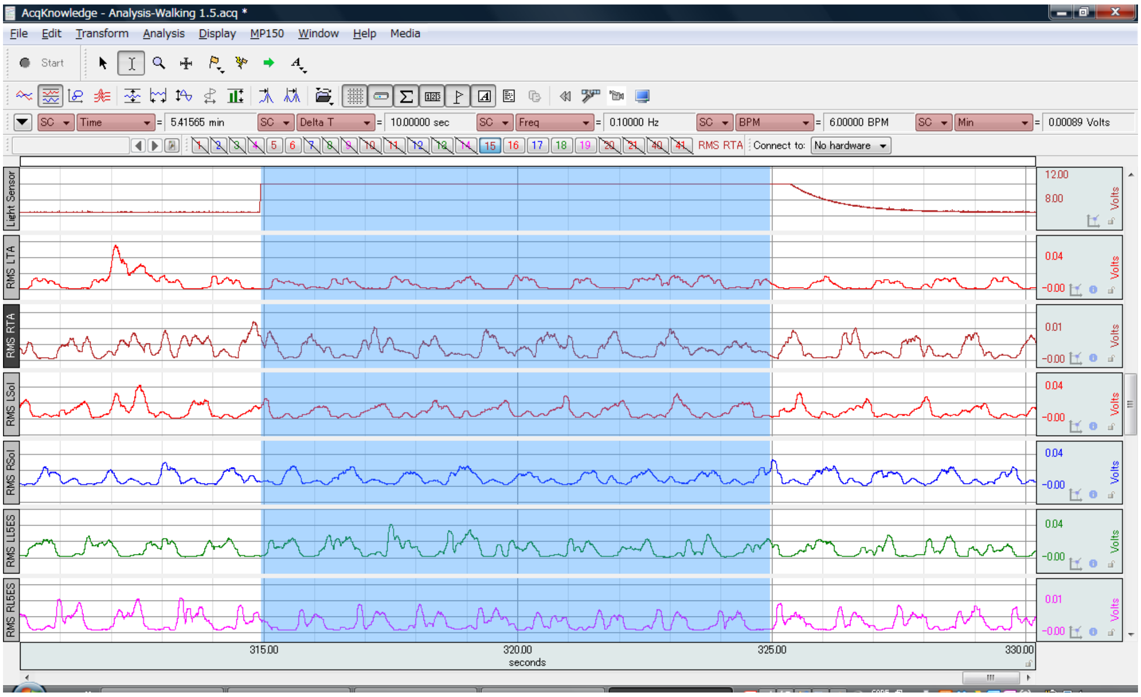

Figure 2-10. RMS of EMG data was selected in 10 s, from the beginning of light sensor recorded the light stimulus to mark the beginning of the trial. The RMS data were obtained after 15-250 band-pass filtering, and derived at a time window of 100 miliseconds.

Application of EMG in ergonomic and physiological measurement

EMG provides easy access to physiological processes that cause the muscle to generate force, produce movement and accomplish various functions which allow humans to interact with the environment (De Luca, 1997). EMG has been proven to be very useful in assessing work positions and activities, usability of design tools and workstation, and in longer term also able to help us to understand better some features of humans physiological retained during evolutionary process. In the study of evolutionary locomotion for example, it was reported that the rhythm of human bipedal gait cycle and phase of ankle flexor and extensor muscle have resemblance to that of quadrupedal animal rhythmical gait, showing the relation of primary and secondary gait in quadrupeds and bipeds (Golubitsky et al., 1999).

2.3 Symmetry Index

The symmetry index developed by Robinson et al. (1987) has been widely used to quantify gait symmetry at discrete time points during stance:

$$\mathrm{SI} = \frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \mathrm{x}100\%$$

where SI is the symmetry index, x_r is the variable recorded for the right leg and x_l the variable recorded for the left leg.

Si is only used for static measurement. For dynamic measurement, cross-correlation function analysis is more appropriate since it can measure continuous signals in a certain time period.

2.4 Cross-Correlation function analysis

As suggested by Wren et al. (2006), in general, during gait, muscle activity can be quantified in terms of EMG amplitude relative to either a maximal voluntary isometric contraction or a reference based on maximum contraction during walking. However, muscle activity is very common inspected in a subjective way by means of visual examination of EMG data. This method has two limitations, such as the comparisons are less objective and the information about the shape of the EMG signals due to changing magnitude across the gait cycle is often being neglected.

Dynamic EMG has been applied in order to record muscle activity during gait, to describe abnormal motor unit firing patterns and to identify muscles for transfer (Perry et al., 1976; Perry and Hoffer, 1977; Kleissen et al., 1998). Recently, dynamic EMG applications have been focused on comparisons of muscle timing and overall muscle activity. Muscle timing can be calculated by identification of the timing of signal onset and cessation (Bogey et al., 1992; Buurke et al., 2004).

Cross-correlation has advantage as a method which is able to compare objectively the timing and shape of muscle activation signals. Our study used cross-correlation to compare EMG as well as pressure sensor data from different walking and manual pushing trials in able bodied adults. Cross-correlation in our study is useful for identifying the left-right difference dynamically in relation with laterality. The advantages of cross-correlation analysis are that it considers shape of the electrical signals either from EMG, force transducer or pressure sensors in addition to timing objectively rather than subjectively (Wren et al., 2006).

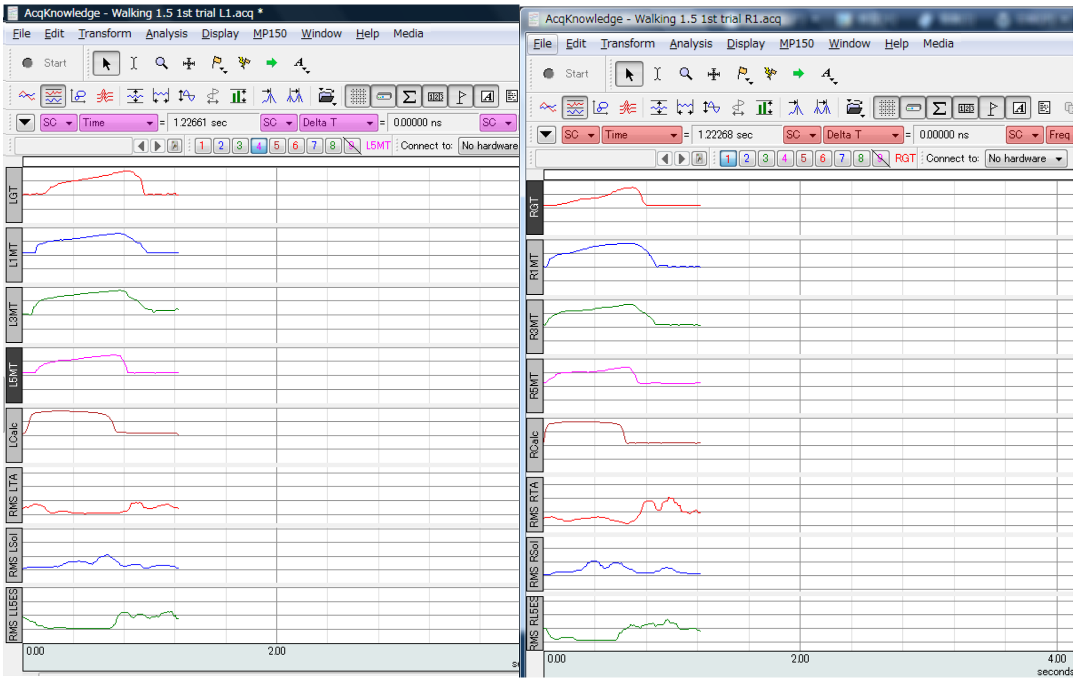

Figure 2-11. Foot pressure and EMG RMS data were divided into each gait cycle and re-sampled into 100 data points in each gait cycle. Left foot pressure sensor and muscle activation data were analysed in the left gait cycle, and the right foot pressure sensor and muscle activation data were analysed in the right gait cycle. For cross-correlation analysis, left and right side data were paired each other.

Cross-correlation involves correlating 2 different time-varying signals, *x(t)* and *y(t),* against each other. Both processes involve iteratively shifting 1 signal forwards and backwards in time against the other signal that is held stationary with a correlation value, *Rxy*, being calculated at each increment. One of the most useful output measures resulting from cross-correlation is the phase delay (τ*). The value of τ* where the maximal correlation occurs has been used in studies for determining muscle onset latencies, order of muscle recruitment, or influences of muscle activation on center of pressure (COP) (Nelson-Wong et al., 2009).

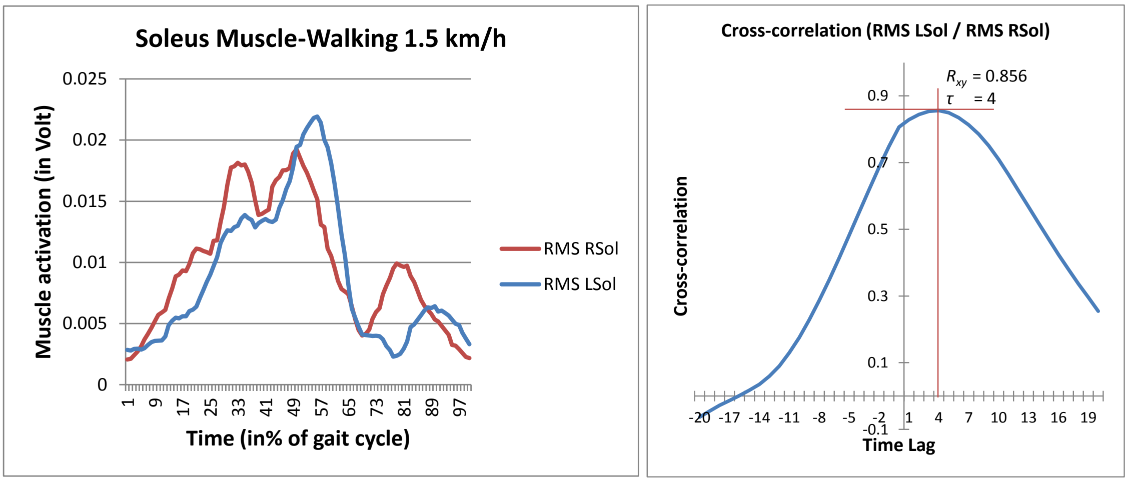

Figure 2-12. Cross-correlation analysis of soleus muscle activation. The cross-correlation coefficient, R_{xy} = 0.856 shows that left and right soleus muscle activation waveforms are highly similar because the value is close to 1, however time lag (τ) = 4, shows that the right soleus muscle activated earlier than the left soleus muscle during their respective gait cycle. Cross-correlation analysis of two signals requires the two signals consist of the same number of data point, thus all data in one gait cycle were re-sampled into 100 data points.

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Chapter 3

The effect of different trunk inclination on bilateral trunk muscular activity, centre of pressure, and force exertions in static pushing postures

3.1 Introduction

Manual pushing is very common in almost every sector of human life. In hospitals, food and supplies are delivered with trolleys (Retsas and Pinakahana, 2000; Das et al. 2002; Hartwell and Edwards, 2003; Pietersma et al. 2003) as well as moving patients on a stretcher or wheelchair (Hignett and Richardson, 1995; Retsas and Pinikahana, 2000; Saltzherr et al. 2008; Sebbane et al. 2012). In passenger aircrafts, food and beverages are served on a trolley (Jaeger et al. 2007; Glitsch et al. 2007). In factories, materials are also moved on trolleys (Okunribido and Haslegrave, 2003). In construction (van der Molen et al. 2011) as well as agriculture, manual pushing activity is performed extensively. In general, 50% of industrial manual materials handling involves pushing and pulling (Granata and Bennett, 2005). Most products that are too heavy to be lifted by one person usually have certain mechanisms to allow them to be pushed. In sports like sumo (Umeda et al. 2008) and rugby (Wu et al. 2007), pushing is also common. Some researchers also reported about ipsilateral pushing in stroke patients that is related to symptom assessment and rehabilitation (Pedersen et al. 1996).

Pushing is an exertion of a force onto an object, in which the direction of the resultant force is anterior to the person (Hoozemans, et al., 1998; Chow, 2010). There are two types of pushing: static pushing that is standing while pushing, where the pushed object is static, and dynamic pushing that is walking while pushing where the pushed object is moving (Todd, 2005; Hoozemans, et al., 2007; Chow, 2010). The initial studies on pushing were based on static horizontal force exertion in the sagittal plane (Resnick and Chaffin, 1995) representing the maximum isometric force (Mittal, et al., 1997), with most studies addressing handle height, especially at hip, elbow and shoulder height (Resnick and Chaffin, 1995; Jansen, et al., 2002; Hoozemans, et al., 2007). Most previous research studied the upper extremities, especially the hand, as the contact area, because of the assumption made by Metter et al. (1997) that the combination of eight forces exerted by the arm and two gripping forces are representative of total body strength (Voorbij and Steenbekkers, 2001).

The human body like other organisms, has been evolved for millions of years to be able to perform the maintenance of posture and locomotion for survival, however the industrial revolution which has changed human daily rhythm induced higher possibility of low back pain to bipedal humans (Kumar, 2002; Schilling et al. 2005). Pushing was found to be associated with musculoskeletal complaints, in particular the shoulder, forearm, and lower back areas (Jansen et al., 2002; Bao and Silverstein, 2005; Theado, et al., 2007; Hoozemans, et al., 2007). However, the extent to which the likelihood of musculoskeletal injury in the lower back, shoulder, and forearm was affected by peaks in pushing force remains unclear (van der Beek et al., 1999). Different findings on optimum handle height were also reported (Lee, 1991; Resnick and Chaffin, 1995; Hoozemans, et al., 2007). Pushing is a total body effort, and thus research on the lower extremities is required (Voorbij and Steenbekkers, 2001), and further research should also explore trunk muscular activity in relation to lower back injury (Hoozemans, et al., 2007). Todd (2005) suggested that the high force requirement and poor working postures adopted by workers during pushing are likely to lead to an increase in musculoskeletal injuries.

Pushing forces exerted in dynamic pushing are divided into three phases: the initial phase, sustained phase, and final phase (Jansen et al., 2002; Hoozemans et al., 2007). During the initial phase, the initial peak force is exerted to accelerate, followed by a smaller sustained force to maintain the movement, and finally the ending phase which actually involves a pulling force (van der Beek et al., 1999). A static pushing study was only able to represent dynamic pushing at low velocity (Resnick and Chaffin, 1995). Isometric strength based on isometric muscular activity during static pushing represents the absolute maximum force a subject can exert, while isokinetic pushing strength, which has a value between 8 to 25% lower than isometric pushing strength, is based on movement and examined during dynamic pushing (Kumar. 1995; Mittal et al., 1997; Chow, 2010). Human strength evaluation is necessary to establish population norms for better design of tools (Chaffin, 1975). Furthermore, isometric data have been the reference for ergonomic guidelines for a long time because they provide information regarding fatigue, productivity, discomfort, and the risk of suffering disorders based on a function of muscle voluntary contraction and time (Mamaghani, 2002). Static pushing can be assumed as having a closer relation to the initial phase in dynamic pushing and also has importance as a reference, because in static pushing the variables are confined to internal exposure measures (forces and muscular activity within the human body), while external exposures (intensity-distance, frequency of pushing over time, environment, etc.) are not measured.

Pushing postures such as trunk orientation and the calculation of shoulder moments during hand-force exertion have become references for lower-back loading assessment using digital human models (Hoffman, et al., 2007). However these models often ignored the physiological aspects and muscular activity that are not solely influenced linearly by body postures during task execution. While trying to push very hard, the subject generally does not stand erect, but rather inclines his body and bends his legs, trunk, and arms (Kroemer, 1969), therefore, in this study we tried to reproduce these body positions in various static pushing postures. A proper static pushing posture should be characterized by harmony among greater pushing force exertion, acceptable muscular activity, and good balance control. The objective of this study is to explore the influence of body inclination on the activation of trunk muscles and on maximum pushing forces, vertical ground reaction forces (GRF), and the standard deviation (SD) of centre of pressure (COP) displacement to identify recommendations for good body posture in static pushing. The methodology also used produce data that show symmetry/asymmetry in manual pushing, which to our knowledge, have been given less attention in previous studies.

3.2 Methods

3.2.1 Subjects

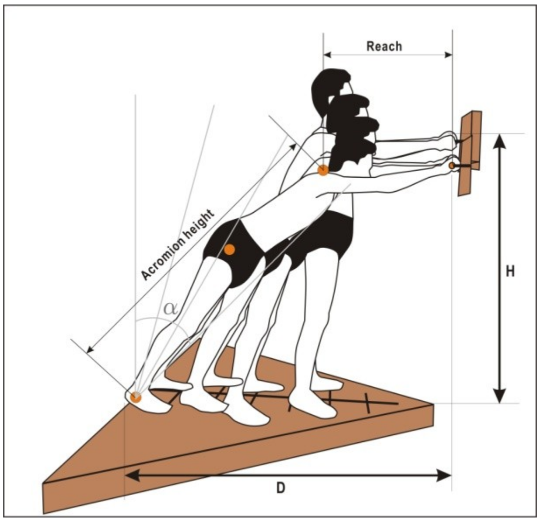
Ten healthy male Asian subjects (age 29 $\pm$ 5 years, range 21-37 years; height 169.0 $\pm$ 5.7 cm, range 160.5-180.0 cm; weight 64.7 $\pm$9.7 kg, range 48.0-82.5 kg; BMI 22.7 $\pm$ 3.0, range 16.8-27.6) volunteered for the experiment. All subjects had no recent history of musculoskeletal complaints. Subjects' laterality data were assessed by Waterloo Footedness Questionnaire (WFQ) and Waterloo Handedness Questionnaire (WHQ) (Elias, et al. 1998). The results showed that all subjects were right handed (+49.6$\pm$13.4) and right footed (+9.8$\pm$4.4).The study was approved by Research Ethics Committee of the Chiba University Faculty of Engineering (24-34). Prior to the experiment, each subject had given informed consent.

3.2.2 Instruments

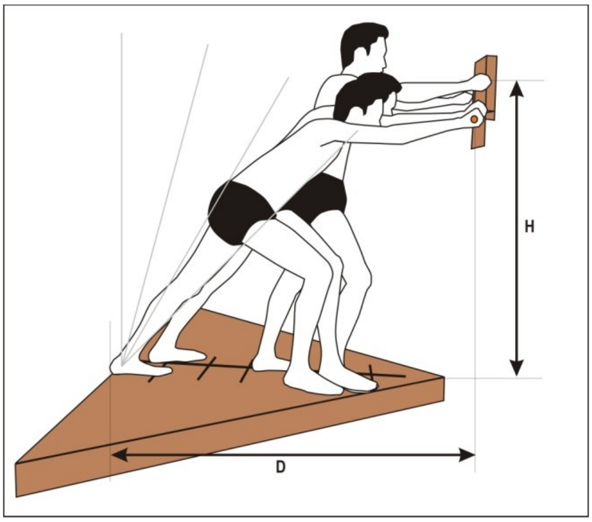
As shown in Figure 3-2, two force plates were used. One is a ground force plate in the shape of an equilateral triangle with a side length of 1790 mm. It contains a Kyowa LCN-A-1 KN (Kyowa Corp., Japan) load-cell unit at each corner with a load-cell to load-cell distance of 1530 mm, and each load cell was connected to a WGA-100B-11 (Kyowa Corp., Japan) strain amplifier which has only one input channel. The other one is wall force plate made in the shape of a square with a side length of 400 mm. It also contains one Kyowa LMB-A-500 N (Kyowa Corp., Japan) load-cell unit at each corner with a load-cell to load-cell distance of 300 mm and all the four load cells were connected to a TEAC SA-30A (TEAC Corp., Japan) strain amplifier with four input channels. All of the strain amplifiers were connected to a Biopac MP 150 (Biopac Systems, USA) data acquisition system and a personal computer for data recording. The wall force plate and the ground force plate are used to measure pushing force and GRF in vertical direction respectively, and both force plates also measured COP which has been widely used in the last few decades as an indication of postural stability (Sriwarno, et al. 2008; Takeuchi, et al. 2009; Ruhe, et al. 2011).

The electromyogram (EMG) electrodes used were Ag/AgCl TSD150A surface EMG electrodes connected to another Biopac MP 150 data acquisition system. EMG data were collected bilaterally from back muscles including the following: latissimus dorsi (LD) sites were lateral to the T9 point over the muscle belly on the line drawn from the acromion to the iliac crest; thoracic erector spinae (T9ES) sites were about 3-5 cm lateral to the T9 spinous process; lumbar erector spinae (L5ES) sites were about 2 cm lateral to the L5 spinous process. Data was also collected from the right side of abdominal muscles as follows: the upper rectus abdominis (URA) site was in the approximate centre of the second uppermost section of the muscle belly; the lower rectus abdominis (LRA) site was in the approximate centre of the lowermost section of the muscle belly about 2 cm lateral to the umbilicus; the external oblique (EO) site was approximately 10-15 cm lateral to the umbilicus on the line from the lowest rib to the iliac crest; the internal oblique (IO) site was between the anterior superior iliac spine and midline slightly superior to the inguinal ligament. To identify the muscle sites manually, palpation methods (Muscolino, 2009) combined with landmark identification and pen marking on the trunk were performed. EMG normalization was adapted from muscle testing methods by Vera Garcia, et al., (2010) and Kendall, et al., (1983) as follows: latissimus dorsi was tested by performing hold adduction; thoracic and lumbar erector spinae were tested when subjects laid prone and extended their trunk against gravity; upper and lower rectus abdominis were tested when subjects sat up and held position at halfway between lying supine and sitting; internal oblique and external oblique were tested when subjects sat up combined with performing trunk flexion and rotation. Each muscle was tested in three trials for 3 seconds with 1 second holding position to achieve MVC, a 5 minutes rest between trials was given, and the maximum muscular contraction value achieved was used as 100% MVC reference subtracted by basic noise level which is used as 0% reference. MVC trials and pushing tests are assigned to different sessions in order to avoid fatigue.

3.2.3 Experimental procedures



(a)



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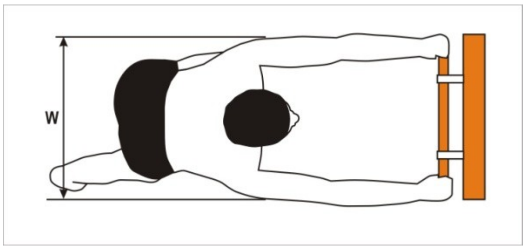

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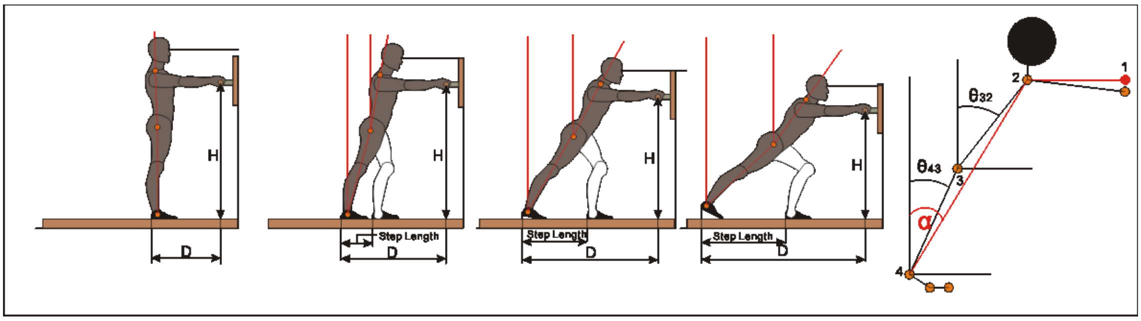


(d)

Figure 3-1. Pushing postures in four body inclinations: (a) 0°, 15°, 30°, and 45°, with parallel feet; and (b) 15°, 30°, and 45° with staggered feet (right foot in front) (α =inclination angle; D= horizontal distance from heel of posterior foot to wall force plate handlebar; H= vertical distance from ground to acromion); (c) W= step width according to shoulder width, (d) step length measurement and the comparison of idealistic inclination of α and the actual inclination of trunk (θ_{32}) and lower limbs (θ_{43}) against vertical line.

Figure 3-1 shows how static pushing trials were performed in four inclinations: 0^o (erect standing), 15^o, 30^o, and 45^o from the vertical line. Erect standing (0^o) was performed with parallel feet (PF) only, while the others were with PF, staggered feet right foot in front (SFR) and staggered feet left foot in front (SFL). Trials were performed with fully extended arms, the shoulder was at around the same level as the wall force plate handlebar, and the lower limbs opened laterally as wide as shoulder width. Trials were performed three times, for a duration of 5 seconds each time with 1-minute rests between trials, and 3-minutes rests between feet stances. All trials were performed in randomized order. In stances with staggered feet, the subjects were allowed to choose their step length freely. Subjects were given opportunity to do some practice before the recorded trials so that they could perform the trials without further adaptation. During the experiment, subjects were instructed to repeat the same postures performed during practice, confirmed by inclination and step length measurements. With regard to volar aspect, the contact of all subjects to ground force plate and wall force plate was barehanded contact to wall force plate wooden handlebar, and rubber soles of footwear with ground force plate surface made from plywood which gave enough friction to avoid slip. To ensure that the planned static posture was achieved during the trials, the wall force plate distance from posterior foot was calculated, and the placement was adjusted based on reach and acromion height and on the trigonometric function of each inclination angle intended, as shown in Figure 3-1:

$$D = (\text{Acromion height} * \sin \alpha) + \text{thumb-tip reach} \quad (1)$$

$$H = \text{Acromion height} * \cos \alpha \quad (2)$$

Where:

A = inclination angles (0^o, 15^o, 30^o, 45^o)

D = horizontal distance from the heel of posterior foot to wall force plate handle bar (mm)

H = vertical distance from ground force plate surface to wall force plate handle bar (mm)

Trunk inclination was then confirmed visually using a flexiometer from a distance of approximately 1.5 meter by measuring the angle between vertical line and virtual line connecting markers on lateral malleolus, greater trochanter, and acromion. Markers as references were put on the acromion, greater trochanter, and malleolus according to method by O'Sullivan, et al. (2002). During the changing of the wall force plate heights, due to re-setting, unintended rest periods of 8 minutes were given to the subjects. To control visual influence on balance (Bessou, et al. 1999; Golomer and Mbongo, 2004; Nagano, et al. 2006) all subjects were asked to gaze straight ahead to the wall, with their face without eye movement.

3.2.4 Subjects posture

Table 3-1. The distance from the heel of posterior foot to the handlebar of wall force plate in horizontal and vertical from each subject as described in Figure 1 (in mm).

Distance	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	Mean	SD
D 0	766	762	755	773	740	800	825	705	725	691	754.91	38.97
D 15	1123	1123	1122	1142	1100	1160	1225	1056	1067	1056	1116.45	49.51
D 30	1456	1460	1465	1486	1435	1496	1597	1383	1386	1396	1453.41	61.15
D 45	1742	1748	1759	1781	1723	1784	1917	1664	1660	1688	1742.73	71.90
H 0	1380	1395	1420	1425	1390	1392	1544	1356	1322	1410	1397.00	59.04
H 15	1333	1348	1372	1377	1343	1345	1492	1310	1277	1362	1349.50	57.03
H 30	1195	1208	1230	1234	1204	1205	1337	1174	1145	1221	1209.80	51.13
H 45	976	986	1004	1008	983	984	1092	959	935	997	987.82	41.75

Table 3-1 shows the distance from the handlebar of wall force plate to the heel of each subject's posterior foot. While deviation from the planned inclination was expected (as comparison between α with θ_{32} and θ_{43} in Figure 3-1), this calculation will minimize variation between subjects' postures, therefore independent variable on posture was limited only on inclination angle.

Table 3-2. Actual inclination angle as shown in Figure 1 (Mean±SE)

		PF	SFR	SFL	p (SFR-SFL)
θ_{32}(in ^{o})	0^o	0.2±0.2			
	15^o	22.4±1.3	22.2±2.1	22.8±2.3	p>0.05
	30^o	41.3±2.0	42.2±2.2	39.7±2.9	p>0.05
	45^o	55.0±2.0	56.6±2.9	52.2±2.2	p>0.05
θ_{43} (in ^{o})	0^o	0.2±0.2			
	15^o	14.9±0.8	14.9±0.7	16.4±1.0	p>0.05
	30^o	26.0±0.4	27.1±0.8	27.3±1.0	p>0.05
	45^o	36.0±1.8	34.2±3.1	37.9±2.0	p>0.05

Table 3-3. Step length (Mean±SE)

		SFR	SFL	p (SFR-SFL)
Step length (in % body height)	0^o			
	15^o	24.3±1.7	24.8±1.6	p>0.05
	30^o	29.7±2.2	31.8±2.2	p>0.05
	45^o	37.3±2.2	37.0±2.2	p>0.05

As shown in Table 3-2 and 3-3 respectively, when the subjects performing the static pushing in the same body inclination but different feet stances, there was no significant differences observed either in trunk and lower limbs inclination angles against vertical line or step length (p>0.05), indicating that postures performed were relatively symmetrical between SFR and SFL. In greater inclination angles such as 30^o and 45^o, the lower limb inclination angle which is limited by maximum dorsiflexion of ankle joints and flat surface of ground force plate resulted in smaller inclination angle from the intended angle, and this was compensated by greater trunk inclination angle.

3.2.5 Data collection, analyses and measurements

Force plate and EMG data were recorded at a 1000-Hz sampling rate. The force plate output was normalized into a bodyweight percentage.

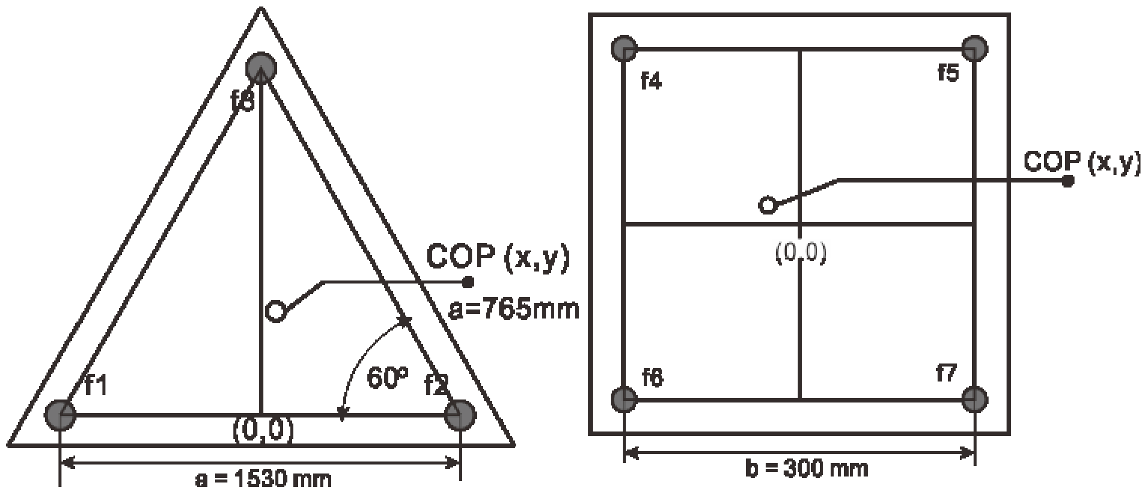


Figure 3-2. Centre of pressure on ground whose (0,0) coordinate was placed on the halfway of the line connecting f1 and f2 (left) and wall force plate whose (0,0) coordinate was placed in the middle of the plane (right).

COP calculation methods were as the following:

Ground COP (x,y) $= \left(\frac{a(f2-f1)}{2(f1+f2+f3)}, \frac{\text{Sin } 60^{\circ} \quad af3}{(f1+f2+f3)}\right)$ (3)

Wall COP (x,y) $= \left(\frac{b[(f5+f7)-(f4+f6)]}{2(f4+f5+f6+f7)}, \frac{b[(f4+f5)-(f6+f7)]}{2(f4+f5+f6+f7)}\right)$ (4)

Where:

a = the distance between load cells on ground force plate, equal to 1530 mm.

b = the distance between load cells on wall force plate, equal to 300 mm.

f1 = vertical force on the first load cell of ground force plate.

f2 = vertical force on the second load cell of ground force plate.

f3 = vertical force on the third load cell of ground force plate.

f4 = horizontal force on the first load cell of wall force plate.

f5 = horizontal force on the second load cell of wall force plate.

f6 = horizontal force on the third load cell of wall force plate.

f7 = horizontal force on the fourth load cell of wall force plate.

f1+f2+f3 = vertical ground reaction force.

f4+f5+f6+f7 = pushing force.

As shown in equation 3 and 4, based on modified method by Hesse et al. (1997), COP was calculated according to the resultants of all the load cells and distance from

(0,0) reference point with assumption that left or below the reference axis is considered negative, and right or above the reference axis is considered positive. SD of COP was based on average of all trials in each subject then averaged for all subjects. On ground force plate, the x axis refers to lateral direction and the y axis refers to antero-posterior direction, while on wall force plate, the x axis refers to lateral direction and the y axis refers to vertical direction.

Raw EMG signals were low-pass filtered with a cut-off frequency of 250 Hz to remove electrocardiogram artifacts, high pass filtered with a cut-off frequency of 15 Hz to remove movement artifacts, and root mean square (RMS) was derived at a time window of 300 milliseconds (Zedka, et al., 1998). By using Fast Fourier Transform analysis, some raw EMG signals with high electrical noise were excluded from further data analyses. As a result, for erector spinae data, only the data from 8 subjects were included.

For measuring symmetry/asymmetry, a Symmetry Index (SI) adapted from equation proposed by Robinson, et al. (1987) was used:

$$\text{Symmetry Index} = \frac{X_R - X_L}{\frac{1}{2}(X_R + X_L)} . 100\% \qquad (5)$$

Where:

X_R = variable measured in the right side.

X_L = variable measured in the left side.

A value of zero corresponds to perfect symmetry between the left and the right side, a positive value indicates the right side magnitude is larger than the left side, and a negative value indicates the left side magnitude is larger than the right side (Herzog et al., 1989). While the original formula applied in force plate data analyses only, in this study, we also compared EMG data.

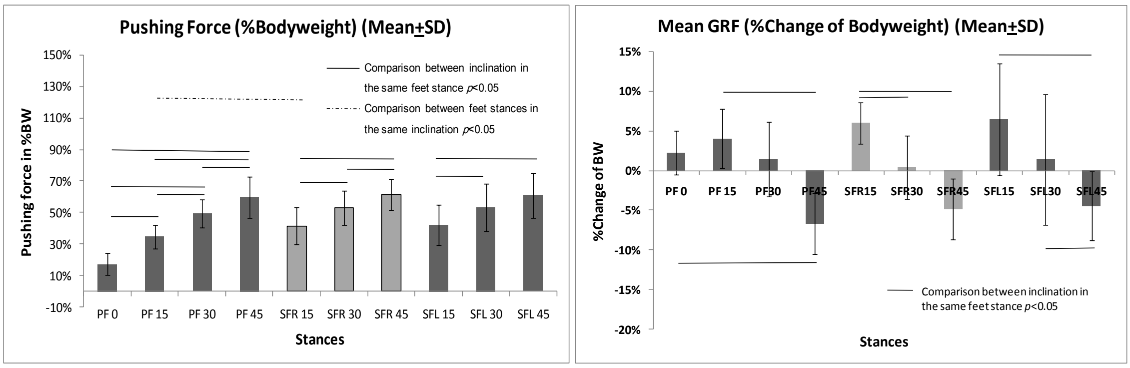
For statistical analysis, the processed EMG and force plate data was inputted into Microsoft Excel 2010J. Further statistical analysis was performed using SPSS 18.0J. The Shapiro-Wilk test was used to determine the probability of normal distribution of the data. Parametric methods used one way ANOVA with repeated measures with the Bonferroni post-hoc test and Pearson’s product moment correlation, while non-parametric methods used the Friedman test with the Wilcoxon signed-rank post-hoc test and Spearman's rank order correlation. The statistical significance level used was $p<0.05$.

To sum up, in the recorded experimental trials, the subjects were instructed to perform various pushing postures, which were restrained by certain different distances between subjects and the wall force plate. For result analysis, the independent variables of this study were body inclination and feet stances (PF, SFR and SFL), while the dependent variables were pushing force, vertical ground reaction force (GRF), standard deviation of both ground and wall force plates COP displacements, trunk muscular activity, and SI of forces and muscular activity.

3.3 Results

3.3 1 Pushing force and ground reaction force

Figure 3-3 (a) shows that the pushing force significantly increased with greater inclination, in both the PF ($p<0.001$) and staggered feet ($p<0.05$) conditions, except between 30^{o} and 45^{o} in the SFL condition ($p=0.06$). Among the different stances, a different pushing force exertion was observed at the 15^{o} inclination, where SFR resulted in a higher pushing force than PF ($p<0.01$), while no significant difference was observed at the 30^{o} and 45^{o} body inclinations ($p>0.05$).



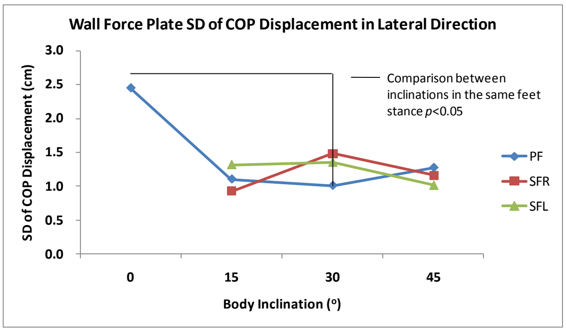
(a) (b)

Figure 3-3. Maximum isometric pushing force given in terms of bodyweight percentage(a) and mean ground reaction force given in terms of percentage of bodyweight change (b).A negative correlation between pushing forces and ground reaction forces was found between the 15^{o} and 45^{o} body inclinations in all stances.

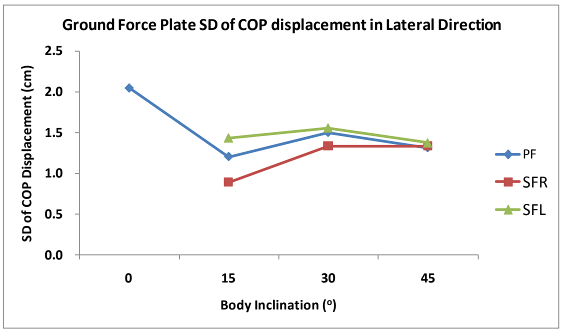
Figure 3-3 (b) shows that the GRF was highest at the 15^o inclination and lowest at 45^o inclination in all stances ($p<0.05$), and that it had a negative correlation with pushing force at inclinations above 15^o in PF (r = -0,527, p = 0.01), SFR (r = -0.485, p = 0.03), and SFL (r = -0.606, p = 0.03). In PF, between the 0^o and 15^o body inclinations, GRF and pushing force had a positive correlation, but this correlation was insignificant ($p>0.05$), suggesting that at body inclinations below 15^o, bodyweight played an insignificant role in determining the pushing force, in contrast with those of between 15^o and 45^o inclination at which bodyweight played a very significant role. In 15^o, 30^o, and 45^o body inclinations, there were no significant GRF difference found in comparison between feet stances in the same body inclination ($p>0.05$).

3.3.2 Wall force plate and ground force plate COP

With regard to the pushing stability on the wall force plate, as shown in Figure 3-4, in all stances, the 30^o inclination resulted in the lowest SD of the COP displacement in the vertical direction especially in the PF and SFL conditions, while no significant difference was found in SFR (Figure 4 (c)). Figure 4 (a) shows that in the lateral direction, no significant difference was found in the SFL and SFR conditions, but in PF the 30^o inclination also resulted in the lowest SD of the COP displacement ($p<0.05$). Erect standing (0^o) significantly produced the highest SD of the COP displacement on the wall force plate both laterally and vertically ($p<0.05$), suggesting that in this posture the subject experienced difficulty in directing the pushing force. Generally, no significant differences were observed in the ground force plate SD of the COP displacement (Figure 4 (b) and (d)).



(a)



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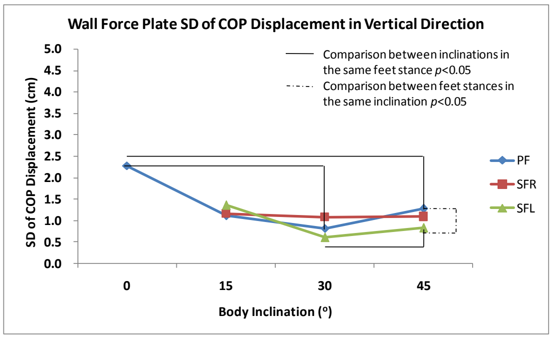

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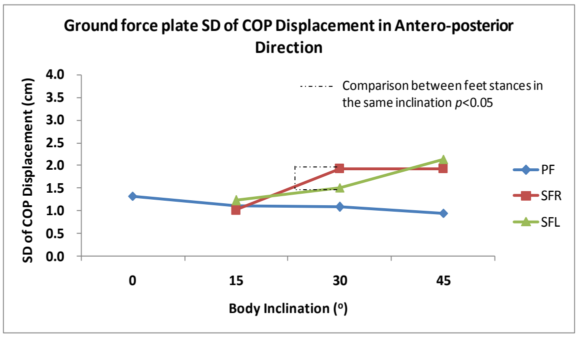


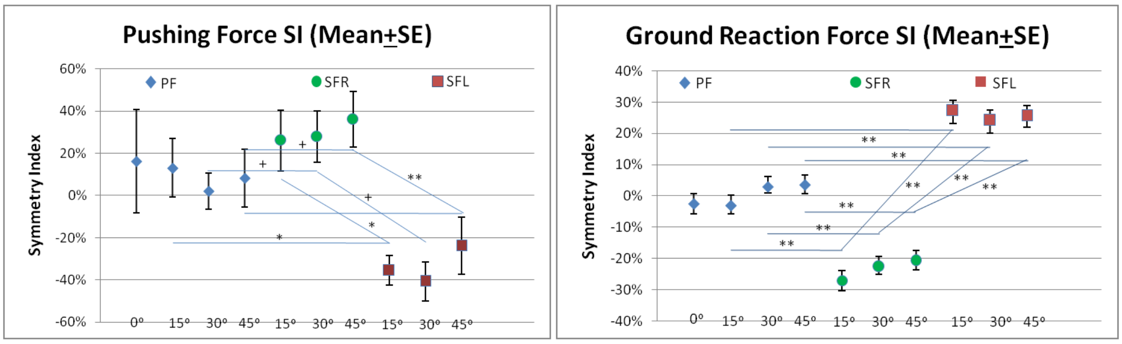
(d)

Figure 3-4. (a) Wall force plate standard deviation of COP displacement in lateral direction, (b) ground force plate standard deviation in lateral direction, (c) wall force plate standard deviation of COP displacement in vertical direction, (d) ground force plate standard deviation of COP displacement in antero-posterior direction. The wall force plate showed more significant differences than the ground force plate.

3.3.3 Pushing force and ground reaction force symmetry index

Regardless of foothold, mean ground COP of each subject was found to be distributed closer to the posterior foot, which in SFR was the left foot and in SFL was the right foot, while in PF the ground COP was evenly distributed between both feet laterally. Figure 3-5(b) shows that in SFR, SI always had negative value, in SFL SI always had positive value, and in PF the value were closer to zero, and SI between the different feet stances in the same body inclination were significantly different (p<0.05).

With regard to pushing force SI, no significant differences due to different body inclinations in the same feet stances were observed, as well as between PF and SFR in the same body inclination (p>0.05) as shown in Figure 3-5 (a). However, SFL SI values were significantly different from both PF and SFR values in 15^{o} and 45^{o} body inclinations ($p<0.05$), whereas in 30° body inclination SFL SI values were found to be significantly different from SFR only ($p<0.05$). The findings that PF SI values were closer to zero, while SFR SI and SFL SI values were positive and negative respectively with different levels of significant differences, suggested a probable influence of laterality on pushing force direction. This finding showed that despite of the absence of lateral postural differences in static pushing, the pushing force direction is likely to be influenced by individual differences, which are difficult to assess solely by using digital simulation human model systems in posture evaluation.



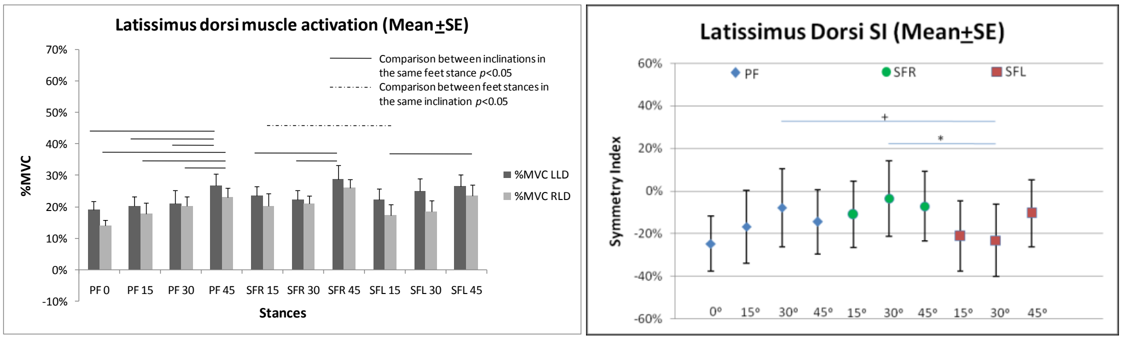
(a) (b)

Figure 3-5. Pushing force SI (a) and ground reaction force SI (b) (**$p<0.01$, *$p<0.05$, $^{+}p<0.1$).

3.3.4 Muscular activity

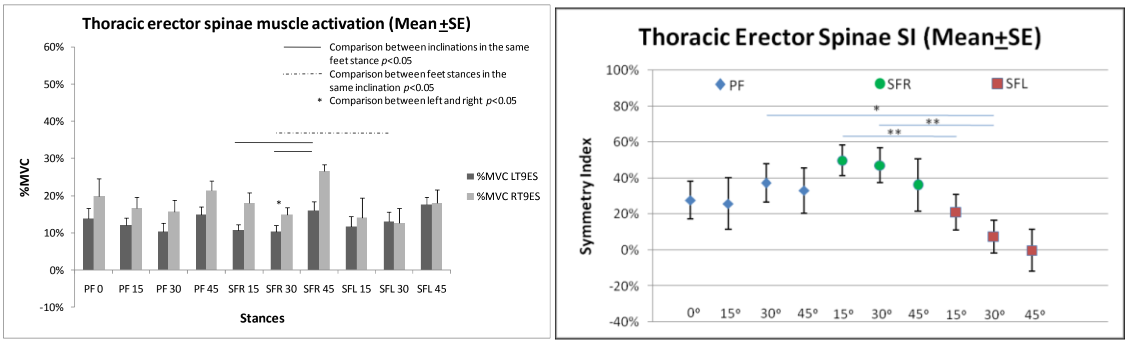
Figure 3-6(a) shows that both left and right LD activation increased significantly with inclination in all stances, with a 45° body inclination generating the highest muscular activity ($p<0.05$), except between 15° and 30° where no significant increase was observed in either SFR or SFL ($p>0.05$). In a comparison between stances at the same body inclination, in general an insignificant difference was found in both left and right LD at all body inclinations ($p>0.05$); except at the 15° inclination, at which SFR generated higher right LD muscular activity than SFL (Z = -2.090, p = 0.037). This result suggested that stances have no influence on LD muscular activity. Between left and right LD no significant differences were found at any of the body inclinations or stances ($p>0.05$), suggesting that LD activation was laterally at about the same level despite the difference in force direction output.

With regard to SI, significant differences were found only in 30° body inclination, where SFR SI was significantly different from SFL (p=0.011), while PF was found to show trend of difference from SFL (p<0.1) (Figure 3-6 (b)). There were no significant differences observed in other comparisons.



(a) (b)

Figure 3-6. Latissimus dorsi muscular activity during static pushing task (a). Latissimus dorsi SI (b) (*p<0.05, ^{+}p<0.1).



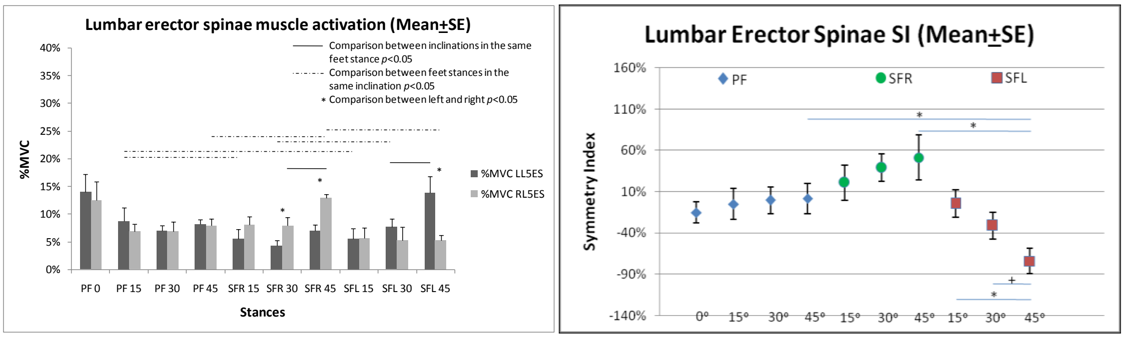
(a) (b)

Figure 3-7. Thoracic erector spinae muscular activity during static pushing task (a). Thoracic erector spinae SI (b) (**p<0.01, *p<0.05)

Figure 3-7(a) shows that T9ES muscular activity did not significantly increase with body inclination (p>0.05), except in the case of the 45° body inclination during SFR, in which the left T9ES was activated significantly more than in the 15° body inclination (Z = -2.803, p<0.01) and 30° body inclination (Z = -2.803, p<0.01). In a comparison between stances at the same body inclination, namely 30° body inclination, the left T9ES during SFL had significantly higher activation than in SFR (Z = -2.701, p =

0.007), although no significant differences were found in the other experimental conditions. In general, no significant differences were observed between the left and right T9ES except at the SFR 30° body inclination, at which the right T9ES had significantly higher activation than the left T9ES (t(8) = -2.470, p = 0.039).

T9ES SI showed that different body inclinations in the same feet stance does not cause significant differences (p>0.05). However, SFL SI were found to be significantly different from SFR in 15° body inclination as well as 30° body inclination (both p=0.003), and also from PF in 30° body inclination (p=0.047).

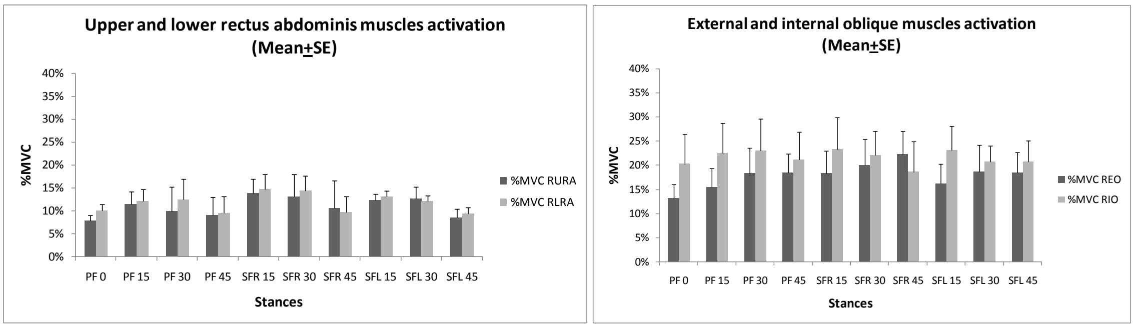


(a) (b)

Figure 3-8. Lumbar erector spinae muscular activity during static pushing task (a). Lumbar erector spinae SI (b) (*p<0.05, ^{+}p<0.1).

In the lower back muscles, Figure 3-8(a) shows that ipsilaterally to the foothold, L5ES muscular activity increased significantly during both SFR and SFL at the 45° inclination (p<0.05). In the PF condition, the largest activation of lumbar erector spinae was observed during erect standing; however, this activation was insignificant (p>0.05). A comparison between stances at the same body inclination showed that at the 15° body inclination, the left L5ES during PF had a significantly higher activation than in SFR (Z = -2.310, p = 0.021) and SFL (Z = -2.547, p = 0.011), while no significant differences were observed in the right L5ES (p>0.05); at the 30° body inclination, the left L5ES in the SFL condition had significantly higher activation than in the SFR condition (p = 0.041), while no significant differences were observed in the right L5ES (p>0.05); and at the 45° body inclination, no significant differences were observed in the left L5ES (p=0.07), while the right L5ES during SFR had significantly higher activation compared to PF (Z = -2.090, p=0.037) and SFL (Z = -2.803, p=0.005). This finding was further supported by a comparison between the left and right L5ES in which, at both the SFR $30^o$and 45^o body inclinations, the right L5ES had a significantly higher activation than the right L5ES (p<0.05), while in the SFL 45^o body inclination the left L5ES had significantly higher activation than the right L5ES (t(8) = 3.137, p=0.014).

The finding of asymmetrical muscular activity in L5ES was further supported by SI change that was seen in SFL. Between 15^o and 45^o body inclinations, SI changed significantly toward negative value (p<0.05) due to asymmetrical muscular activity between left and right L5ES (Figure 3-8 (b)). This was the only significant difference found due to different body inclinations in the same feet stance. In comparison between feet stances in the same body inclination, significant SI differences were observed only in 45^o body inclination, where PF significantly differed from SFL (p=0.018) and SFR significantly differed from SFL (p=0.021).



(a) (b)

Figure 3-9. Upper and lower rectus abdominis muscular activity during static pushing task (a). External and internal oblique muscular activity during static pushing task (b).

No significant differences were found in any abdominal muscles in all trials (p>0.05). As shown in Figure 3-9 (a), the rectus abdominis muscles showed a decreasing trend with greater body inclination, especially in the staggered stances, in opposition to the increasing trend shown in the back muscles. Figure 3-9 (b) shows the insignificant increasing trend of the external and internal oblique muscles with greater body inclination.

3.4 Discussion

Despite that the focus of this study was on trunk inclination, it should be noted the trunk does not generate the power of pushing, but rather becomes rigid to sustain power generated by upper limbs and shoulder girdles as well as to transmit the power generated in the lower limbs and pelvic girdles to the application point of force through upper limbs as effective as possible. Thus, while trying to push very hard, a person generally tries to arrange his posture so that his strongest muscles can be fully activated, his body weight used, and the chain of force vectors from the supporting ground through his body to the point of force application maintained with minimum loss (Kroemer, 1969). The trunk is where most of the human body mass as well as the centre of mass is located, therefore, by controlling the direction of body mass, it helps to generate larger pushing force.

Heavier individuals have an advantage in that they have enough mass to handle large objects and a higher capacity for infrequently materials handling (Mittal et al., 1997), as shown by the negative correlation between GRF and pushing force and the insignificant difference between stances at the same inclination above 15^{o}, representing a higher utilization of bodyweight (Lee et al., 1991; Holbein and Chaffin, 1997 in Hoffman et al., 2007; Hoozemans et al., 2007). Below 15^{o}, the role of bodyweight was insignificant; however, this finding was not reported in previous studies. For more complex operation such as frequently dynamic pushing, greater force exertion due to body mass is accompanied by greater metabolism costs required to perform the task (Lagerloef, 2007; Watson, et al. 2008). Ayoub and McDaniel (1974), and Resnick and Chaffin (1995) found a mean static pushing force of 620 N for young men with the hands at about 80% of shoulder height or about 100 cm above the floor, while Chaffin et al. (1983) found a maximum isometric pushing force at a handle height of 68 cm. In the present study the lowest handle height examined was 93.5 cm at the 45^{o} body inclination. Wu et al. (2007) studied maximum pushing forces among rugby players in 3 body inclinations, 36%, 38%, and 40% of body height, and they reported that maximum pushing forces were found at 40% of body height, suggesting that correlation between body inclination and pushing force is probably positive to a certain point and then turns negative. However this study and the study by Wu et al. (2007) were based on flat ground, where in greater trunk inclination, contact area between footsole of the posterior foot and the ground decreases due to limited ankle dorsiflexion angle. The results can be different in condition where the stiction was large enough, such as the use of starting blocks commonly used in sprint running.

There were no significant maximum pushing forces difference between feet stances in the same body inclination, except in 15^{o}, where SFR resulted in greater pushing force than PF. In general, this finding is in agreement with the study by Wu et al. (2007) who also found that between parallel feet and non-parallel feet there was no significant difference in pushing forces.

The lowest SD of COP displacement was found at the 30^{o} inclination in all foot positions. Previous studies on static pushing by Granata and Bennett (2005) found that the greatest stability during maximum pushing force exertions was in a 40^{o} trunk angle. In the present study, 40^{o} body inclination was not measured, however 45^{o} body inclination was not found to generate good stability, especially on the pushing force. The finding of symmetrical GRF SI is in agreement with the study by McCurdy and Langford (2005) on dominant and non-dominant squat strength, in which no significant difference in strength between the dominant and non-dominant legs was found. Pushing force SI shows no significant difference between PF and SFR, however both feet stances was found significantly different from SFL in both 15^{o} and 45^{o} body inclinations. However, in 30^{o} body inclination, no significant differences were observed between feet stances; this supported the measurement of SD of COP displacement that this body inclination generates good stability in pushing force direction.

Several studies performed on human walking found that lateral balance was more difficult to control than fore-aft balance, as proven by larger variability in step width than step length (Courtine and Schieppati, 2003; Owings and Grabiner, 2003; Kiriyama et al., 2005). Even though humans have a relatively larger sole-area than four-legged animals, bipedal humans modulate our forward movement with only two legs and this functional and anatomical differences probably influence lateral balance (Warabi et al., 2004). In human evolution, from quadrupedal primates into bipedal human, forelimbs tend to support less proportion of body weight and left the role of body support and propulsion to hind limbs (Vilensky and Larson, 1989). In pushing, the condition is reverse to the evolution path especially when greater trunk flexion is required to push a heavier load, where arms should support greater bodyweight. This condition resembles Gray's biomechanical model on quadrupedal animals based on four-legged tables where stability can be well maintained if three feet kept on the ground at same time when the animal is moving slowly (Gray, 1944; Vilensky and Larson, 1989). This study results and that of Granata and Bennett (2005) showed that greater distribution of load among upper and lower limbs generated better stability, at least around 30-40^o of body inclination.

Other aspects that influence lateral balance are laterality and interlimb coordination. Laterality has been associated with cortical representations (Dassonville et al. 1997; Ziemann and Hallett, 2001; Liepert et al. 2001; Lutz et al. 2005; Bernard et al. 2011; van den Berg et al. 2011) and asymmetrical size of corpus callosum (Witelson, 1985; Luders et al 2010). Most laterality studies were based on handedness (Braccini, et al. 2010; Leca, et al. 2010; Hopkins, et al. 2011; Duarte, et al. 2012). The influence of footedness may also exist. Footedness was found to influence balance ability and asymmetrical muscle size (Chibber and Singh, 1970; Schneiders, et al. 2010) as well as the muscular activity that were reported in the lower limbs (Hoshikawa, et al. 2009; Oshita and Yano, 2010; Yoshioka, et al. 2011). In gait studies, a more lateral centre of mass displacement was observed when starting with the dominant foot (Dessery, et al. 2011). In this study, when the non-dominant foot was placed in front during staggered feet stance, lower SD of COP displacement were observed only in 30^o body inclination on antero-posterior direction of ground COP and 45^o body inclination on vertical direction of wall COP. However, different pushing force SI values were observed when the non-dominant foot was in front. These results probably confirm the statement by Hart and Gabbard (1996) that leg choice for bipedal stabilization may be dependent on the nature and complexity of the task. Hip mechanism influence on laterality, especially in a comparison between the torques of hip sides as reported by Neumann et al. (1988) that it is intermediary between upper and lower limbs, as well as mechanism responsible for upper and lower limbs coordination are probably also significant.

Laterality has been less studied in manual materials handling, despite Coren (1992) reporting that left-handers are twice as injury prone than right-handers in various activities such as work and sports. Dessery et al. (2011) concluded that it was a common mistake in ergonomic studies to assume humans bilateral symmetry by focusing on the dominant limb and generalizing the results to the other limb and neglected limb preference. Symmetrical ground COP distribution accompanied by asymmetrical wall COP in this study suggested that the study of body coordination should avoid generalization of bilateral symmetry of the human body.

In pushing, greater trunk flexion is required to push a heavier load. However, the spine is vulnerable to instability during flexion and to maintain trunk stability requires support from combination of trunk muscles with contribution determined by magnitude and direction of loading (Cholewicki and Van Vliet IV, 2002; Granata and Bennett, 2005).

Latissimus dorsi together with teres major, and caudal parts of the pectoral muscles are sometimes referred to collectively as “propulsive muscles of the shoulder” (Larson and Stern Jr. 2007). In this study, latissimus dorsi muscular activity was found to be increased along with inclination regardless of stance. Insignificant differences between the left and right latissimus dorsi muscles showed that the laterality of this muscle has no correlation with pushing direction in the static pushing condition. In comparison with quadrupeds, latissimus dorsi muscular activity shows the extent of the load carried by the forelimb (Larson and Stern Jr. 2007). Hence, greater latissimus dorsi muscular activity along with greater body inclination represents greater bodyweight supported by the arms which is further used as pushing force.

The trunk and hip muscles, especially the erector spinae were found to play an important role in proactive control balance during walking by maintaining upper body steadiness, and they start to become active before progression begins and remain active during the first half of the stance phase (Basmajian, 1975; Thorstensson et al., 1982; Kumar, 2002; Warabi et al., 2004) with only secondary contribution to twisting activities (Kumar, 2002). A previous study reported that during various isometric trunk exertions trials, the spine was most vulnerable to instability during flexion trials, and lumbar erector spinae, an inter-segmental muscle was found to have the greatest effect on spine stability (Cholewicki and Van Vliet IV, 2002). The lumbar erector spinae muscle showed an insignificant decreasing trend along with body inclination in the PF condition. At the 15^{o} inclination, PF showed greater muscular activity than both SFL and SFR. The largest muscular activity of the lumbar erector spinae was found during the ipsilateral staggered feet condition at the 45^{o} inclination, which showed that hip flexed due to stepping was critical in generating contraction. The flexed hip during stepping is also important to maintain balance whereas the extended hip on the contralateral side is important for propulsion. This assumption is supported by asymmetrical activation of the lumbar erector spinae muscle at 45^o body inclinations. While previous studies found that a lower handle height resulted in larger compression forces in the lower back (Lee et al., 1991, Hoozemans et al., 2007), the present study found that flexion due to stepping was actually more critical, since under the PF condition there was no significant increase in lumbar erector spinae muscular activity. The staggered feet postures, with the front foot under the trunk, confirmed a study of static pushing by Hoffman et al. (2007), which suggested that a staggered feet posture is a conservative strategy to maintain balance in the case of the sudden removal of hand force.

With regard to SI, latissimus dorsi SI were always negative contrary to the thoracic erector spinae SI which were always positive, indicates that cross-symmetry happened between the two segments of trunk, namely the middle part and the upper part. That lumbar erector spinae SI in SFL were found to be significantly different from both PF and SFR in 45^o body inclination indicates that in this body inclination asymmetrical muscular activity occurred, and since balance control is influenced by laterality, higher demand of muscular activity for balance control will result in greater muscular activation. To our knowledge, this condition was not described in the previous studies.

Data collection on the abdominal muscles was inhibited by the fact that subjects with higher BMI, especially with BMI>22, which covered up to 60% of the subjects. These subjects showed insignificant changes in abdominal muscular activity, which confirmed the difficulty of measuring abdominal EMG among individuals with high BMI and high subcutaneous fat in accordance with a previous study by Nordander et al. (2003).

Both from the perspective of stability and lumbar erector spinae contraction, 30^o body inclination produced relatively low deviation and low muscular activity, suggesting that it is probably the most optimum inclination. In static pushing or pushing a heavy load during slow movement, 30^o body inclination resulted in better distribution of bodyweight among upper and lower limbs, and gave good balance control, while still generate great pushing force exertion without significant increases in lower back muscular activity.

The limitation of this study is that we enforced strict control on subjects angle joints to achieve the intended inclination angles to measure, and despite enough practice being given for each subjects to perform their maximum pushing force in the four trunk inclination angles, subjects were not given freedom to exercise their best postures to achieve their real maximum pushing force beyond the four body inclinations. There is possibility that their best postures require trunk inclination angle outside of the measured ones and more varied joints angles. However, the procedure also keeps the results pure from other factors besides trunk inclination angles and feet stances, that could complicate analysis. This strict procedure was used since as mentioned in the introduction part, we have the assumption that the various and contradictory results of previous studies were due to a less controlled experiment where subjects could perform the trials freely, thus inter-subject variability became larger.

3.5 Conclusions

It can be assumed that a greater pushing load requires a greater pushing force, which in turn is enhanced by the utilization of bodyweight through body inclination. As the body inclines lower, the step length that is required to maintain balance becomes longer; based on proximity to the centre mass of the body, greater hip flexion occurs, which in turn causes greater activation of the lower back muscles. The pushing strategy differed between below 15^o and above 15^o. While the former depended more on the stance, the latter depended more on bodyweight utilization; therefore, the larger the inclination is, the greater the pushing force became. The optimum stability during pushing was found to occur at the 30^o body inclination. A body inclination of 45^o resulted in the highest muscular activity during the staggered feet conditions, and the difference between this activation and the others was significant. Based on the pushing force exerted, stability and lower back muscular activity, the 30^o body inclination was found to be the most optimum posture for maximum pushing force exertion.

Static testing was only able to represent dynamic testing at a low velocity (Resnick and Chaffin, 1995), therefore, further research on dynamic pushing, which represents the majority of daily pushing activity, is also necessary. Studies of isometric force exertion and muscular activity should also be accompanied by studies of isometric endurance time, as suggested by Mamaghani (2002), in order to be able to predict the ergonomic requirements of manual pushing more comprehensively.

Despite manual pushing having been studied extensively, generalization based on ergonomic principles without consideration of the nature of human physiological conditions will always generate variable results. This study's results also show the necessity of studying the comparison between walking and manual pushing from the laterality point of view.

3.6 References

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Chapter 4

The influence of laterality to different patterns of asymmetrical foot pressure and muscle activation during gait cycle in manual pushing

4.1 Introduction

In pushing, unlike lifting, the load is actually supported by the floor instead of the body and force is exerted to move the object (Mittal et al., 1997). Previous studies on manual pushing were focused on industrial ergonomics to reduce the musculoskeletal injury risk by investigating handle height (Resnick and Chaffin, 1995; Jansen et al., 2002; Hoozemans et al., 2007), upper extremities (Voorbij and Steenbekkers, 2001) and trunk muscles load (Hoffman et al., 2007). Pushing in daily life is also performed in various activities, such as babies perform pushing during learning to walk (Trettien, 1900), people with balance deficiency (Palisano et al., 2007) or older adults use walking aid which involve pushing. Pushing is also common in sports (Wu et al., 2007; Umeda et al., 2008), and in therapy (Pedersen et al., 1996).

In spite of the extensive studies on manual pushing, our literature searching found that gait cycle and laterality have been given less attention. Gait cycle is described as the duration between initial contact of one foot to the following contact of the same foot (Ounpuu, 1994), which includes stance phase and swing phase. Laterality refers to asymmetrical preferential use of limbs and sensory (Schneiders et al. 2010). Dessery et al. (2011) reported common mistakes in ergonomic studies to focus on the dominant limb due to assumption of humans' bilateral symmetry. In gait studies, the symmetry assumption is aimed to ease data collection and analysis (Sadeghi et al. 2000) however this assumption may miss important responses of locomotory system (Haddad et al. 2006).

It is generally understood that about 85% of human population are right-handed (Uomini, 2009). Other suggested around 90% of humans are right-handed and 80% are right-footed (Carey et al., 2001). Based on population study, left-handers who are living in a world designed primarily for right-handers were found to be more accident prone than right-handers (Coren, 1992). Laterality is not always constant from birth, as beside cultural limitation, handedness transfer due to pathological cause was also reported (Jones, 1870) hence, about 10% of each left-handers and right-handers were found switched into the opposite handedness (Coren, 1992).

While many studies have been conducted on handedness, other laterality measures such as footedness have been given less attention despite ear preference was found to be more related to footedness than handedness (Elias et al., 1998; Chapman et al., 1987; Peters et al., 1988; Schneiders et al., 2010). There is no correlation between handedness and footedness was observed (Chhibber and Singh, 1970), and while right foot tend to be the preferred foot of right-handers, the situation remains unclear for left-handers (Chapman et al., 1987; Peters, 1988). In unipedal postural control, difference due to footedness was reported (Golomer and Mbongo, 2004). Dessery et al. (2011) found asymmetrical body motion influenced by footedness in gait initiation. On the contrary, other studies reported insignificant effect of footedness during walking (Zverev, 2006) and gait initiation (Hesse et al., 1997). Sadeghi et al. (2000) formulated a hypothesis that asymmetrical lower limbs behaviour is in existence because of functional differences in propulsion and control. On the other hand, Hart and Gabbard (1997) suggested that in lower limb choice for postural stabilization is task-dependent. These contradictory results were attributed to limitation of methods of previous studies which mostly only involved right-lateralised subjects.

While laterality is evidenced to be an important feature in human life which has social, educational, and psychological implications and affects health, well-being and even life span- such as left-handers were found to have greater risk of injury in various activities-such as sport, working, driving, etc., with risk between 20 to 80% higher than right-handers- as well as shorter life expectancy (Coren, 1992), it is still not well understood, and in manual pushing has been given less attention. Thus, the aim of this study was to investigate the influence of laterality in various velocities of gait cycle on manual pushing by measuring variables such as pushing force, muscular activity and foot contact duration in order to confirm whether different characteristics of right-handers and left-handers exists in manual pushing.

4.2 Methods

4.2.1 Subjects

Subjects were 17 healthy young adult males selected from Chiba University student population (age 28$\pm$5 years; height 169.9$\pm$6.9 cm; weight 64.6$\pm$7.3 kg). Subjects' laterality was measured with Waterloo Footedness Questionnaire (WFQ) and Waterloo Handedness Questionnaire (WHQ) (Elias, et al. 1998). The results showed that 11 subjects were found to be both right-handed (50.25$\pm$12.42) and right-footed (9.25$\pm$4.13), and 6 subjects were grouped into left-handers (-15.17±13.7) and mixed-footers (0.0±7.5). All right-handers were selected with simple random sampling method however left-handers were targeted specifically from observed left-handers group since their population is very small and consistent left-handers are even more difficult to find (Witelson, 1985). The number of 6 left-handers which was the maximum number obtained from the observed population is considered enough since it has been used in previous study (Tan, 1989) and it is enough to examine statistical significance from both groups of handedness based on physiological measurements. Prior to the experiment, subjects were given information on the experiment procedure and they gave signed informed consent which was approved by the Ethics Committee of the Graduate School of Engineering, Chiba University (25-25).

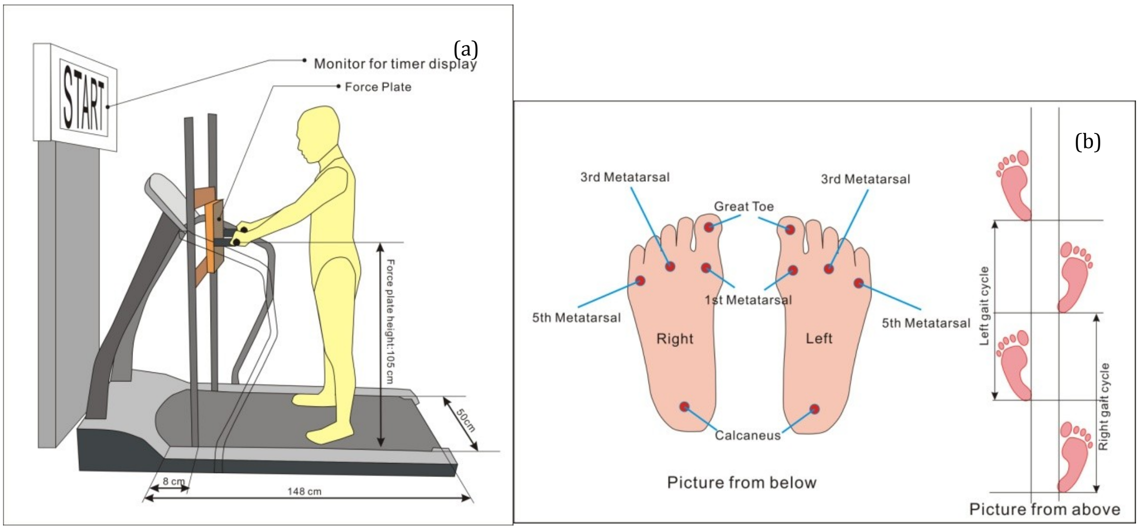


Figure 4-1. Experimental set up: (a) subject pushed a force plate on a treadmill; (b) foot pressure sensors attachment on footsole and gait cycle duration.

4.2.2 Instruments

As shown in Figure 4-1(a), subjects exerted pushing force while walking on a motor treadmill (SportsArt Fitness, Taiwan). The treadmill has overall dimension of 194 cm (length) x 73.5 cm (width) x 150 cm (height) and walking lane of 148 cm (length) x 50 cm (width). Pushing force was measured with wall force plate which contains four LMB-A-500 N (Kyowa Corp., Japan) load-cells, with a load-cell to load-cell distance of 300 mm and all load-cells were connected to an SA-30A (TEAC Corp., Japan) strain amplifier. Force plate handle was placed at 105 cm above the ground, a height which is considered able to accommodate a range of subjects with various body heights (Lee et al., 1991; Resnick and Chaffin, 1995). EMG data were collected bilaterally from the following muscles: tibialis anterior (TA), soleus (Sol), lumbar erector spinae (L5ES), and triceps brachii (Triceps). Both the strain amplifier output and EMG electrodes were connected to an MP 150 data acquisition system (Biopac Systems, USA), whose output was then connected to a personal computer. All trials were recorded using two digital cameras (Canon Corp., Japan) placed at the left side and behind of subjects, to confirm visually post-experiment that no awkward posture was performed during the experiment. In case such a posture was performed, the data will not be processed for further analyses. To simplify the visual confirmation, ball markers were attached on acromion, elbow, greater trochanter, knee and lateral malleolus. In this study, however, we do not perform kinematic analysis.

As shown in Figure 4-1(b) (left section-picture from below), to measure footsole contact timing with the ground, ten FSR-400 pressure sensors (Interlink Electronics, USA) were attached bilaterally on great toe (left: LT-right: RT), 1st metatarsal (left: L1MT-right: R1MT), 3rd metatarsal (left: L3MT-right: R3MT), 5th metatarsal (left: L5MT-right: R5MT), and calcaneus (left: LC-right: RC) based on method used by Kiriyama et al. (2005). All subjects wore the same footwear. Trial duration was marked by visual display timer and light sensor (Kodenshi Corp., Japan).

4.2.3 Experiment Procedure

Subjects performed pushing trials while walking on a treadmill in three velocities: 1.5 km/h (P1.5), 3 km/h (P3), and 4 km/h (P4) (Figure 1(a)). The slope angle of the treadmill was set at 0^{o} (flat surface). The experiment was performed inside a climate chamber with controlled temperature at 26^{o}C as well as controlled lighting. During main trials, subjects were instructed to push at around 50% of their maximum static pushing force. For each condition, trials were performed in 10 seconds for three times with a three minutes rest was given between trials. From the 10 seconds measured data, 3 gait cycles in the middle which is considered more stable were processed for further analyses. All trials were randomised. During trial, subjects were instructed to gaze at the monitor which showed both timer and pushing force feedback. This gazing is advantageous to avoid asymmetrical visual influence on balance control as reported in previous studies (Bessou et al., 1999; Golomer and Mbongo, 2004; Nagano et al., 2006).

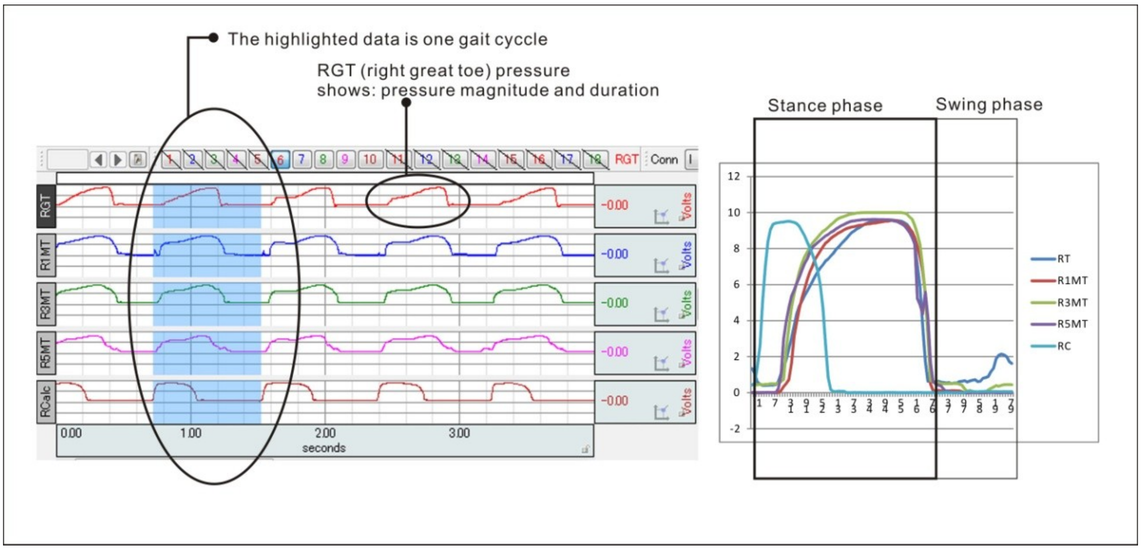


Figure 4-2. Data analysis of gait cycle duration from raw data in acknowledge into processed graph in Microsoft excel.

4.2.4 Data Analysis

As shown in Figure 4-1(b) (right section-picture from above), data analysis was based on one gait cycle, between one heel-strike to the next heel-strike of the same foot, marked by calcaneus foot pressure sensor. Figure 4-2 shows the data processing from raw data in acknowledge software (Biopac systems, USA) into Microsoft excel (Microsoft Corp., USA). Stance phase duration of one gait cycle was a combination of the five pressure sensors attached on footsole (Figure 2 right section). Each pressure sensor showed pressure magnitude and duration of footsole point. We divided data based on right and left side gait cycle, which meant that data from left side muscles and pressure sensors were analysed according to left foot gait cycle, and data from the right side were analysed according to right foot gait cycle. Force plate, EMG and pressure sensor data were collected at a 1000-Hz sampling rate. Raw EMG signals were band-pass filtered between 15-250Hz, and root mean square (RMS) was derived. All data were normalized into 100 data points during one gait cycle.

To measure temporal change in symmetry of variables measured on the left and right side, cross-correlation function (CCF) was used. The coefficients (R_{xy}) vary between -1 and +1, where a positive correlation value indicates the time-varying signals are in phase or increasing and decreasing together, and a negative value indicates inverse relationship (Nelson-Wong et al., 2009). CCF also measured time lag (τ) between two signals. Data measured from the left side during left foot gait cycle were used as template and data from the right side during right foot gait cycle were used as data. If the peak of the right side data were earlier than the left side, time lag will be positive, while if it were later, time lag will be negative.

In further statistical analyses, subjects were grouped into right-handers (RH) and left-handers (LH) groups, since we want to contrast between right-handed and right-footed subjects who are majority of the population with other group of laterality. We analysed the characteristics of each group, and then for CCF and time lag data we also performed comparison between groups. The Shapiro-Wilk test was used to determine the probability normal distribution of the data. To compare data between left and right gait cycle, we employed Student`s paired T-test for parametric method and Wilcoxon signed-rank test for non-parametric method. Data comparison between different velocities in the same group was performed using one-way repeated measures ANOVA with Bonferroni post-hoc test for parametric method and Friedman Test with the Wilcoxon signed-rank post-hoc test for non-parametric method. Because the two groups of subjects have unequal samples, in order to compare both groups, independent T-test for parametric method and Mann-Whitney U test for non-parametric method were used. Statistical significance was set at $p<0.05$. All statistical analyses were performed with Microsoft Excel 2010J and IBM SPSS 17J.

4.3 Results

4.3.1 Subjects anthropometric asymmetry

We recorded anthropometry data bilaterally. LH showed significant differences ($p<0.05$) between left and right side of trochanteric height (left: 86.8$\pm$4.9 cm and right: 86.1$\pm$4.8 cm), upper arm circumference (left: 25.8$\pm$1.8 cm and right: 25.4 $\pm$ 1.9 cm), and lower thigh circumference (left: 39.4$\pm$3.1 cm and right: 39.0$\pm$3.4 cm). RH showed significant differences of acromion height standing (left: 138.5$\pm$7.5 cm and right: 137.7$\pm$7.1 cm), upper arm circumference (left: 26.3$\pm$1.4 cm and right: 26.6$\pm$1.2 cm), forearm circumference (left: 24.7±1.0 cm and right: 25.2±0.9 cm), and lower thigh circumference (left: 40.8±2.7 cm and right: 41.5±3.2 cm).

4.3.2 Gait Cycle Duration

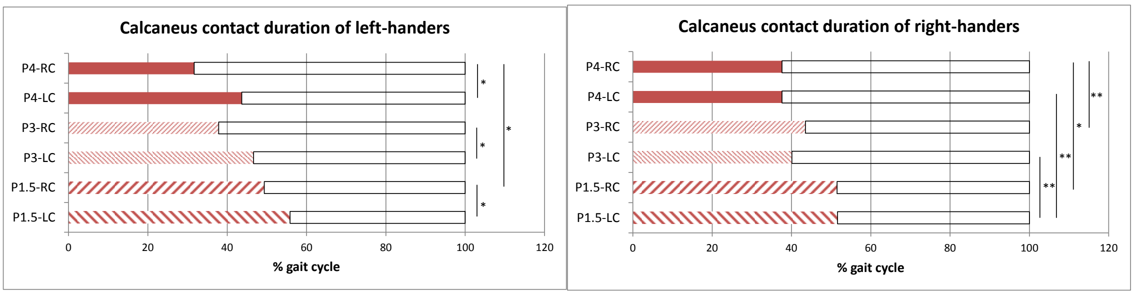
Gait cycle duration was found to be shorter with faster gait during manual pushing in both left and right gait cycle of both handedness groups ($p<0.05$) (Table 4-1). However, asymmetry in gait cycle duration in both groups was not observed ($p>0.05$).

Table 4-1. Gait cycle duration of both left-handers and right-handers (mean ± SD). There was no asymmetry observed on gait cycle duration.

	LH gait cycle duration (in second)			RH gait cycle time (in second)		
Velocity	Left gait cycle	Right gait cycle	*p*	Left gait cycle	Right gait cycle	*p*
1.5 km/h	1.60 ± 0.35s	1.59 ± 0.35s	0.21	1.49 ± 0.22s	1.50 ± 0.23s	0.28
3 km/h	1.14 ± 0.12s	1.13 ± 0.13s	0.27	1.10 ± 0.12s	1.10 ± 0.13s	0.48
4 km/h	0.94 ± 0.09s	0.95 ± 0.09s	0.45	0.96 ± 0.11s	0.96 ± 0.12s	0.35

4.3.3 Foot Contact Duration

Figure 4-3 shows calcaneus contact duration of both group of handedness. During manual pushing at 1.5 km/h, LH only showed asymmetry in calcaneus contact duration with the ground where LC contacted the ground longer than RC ($p<0.05$). LH showed asymmetry in 3rd metatarsal where R3MT significantly has longer contact duration ($p<0.05$), and in calcaneus where LC has longer contact duration ($p<0.05$) at 3 km/h, while at 4 km/h, LH showed asymmetry in calcaneus where LC has longer contact duration than RC ($p<0.05$). Statistical analyses on the influence of velocity found significant influence in RC of LH where 4 km/h velocity shortened contact duration significantly compared to 1.5 km/h ($p<0.05$).



P1.5 : Pushing at 1.5 km/h	RC: right calcaneus
P3 : Pushing at 3 km/h	LC: left calcaneus
P4 : Pushing at 4 km/h	

Figure 4-3. Calcaneus contact duration of LH (left) and RH (right). Filled area represents contact duration of calcaneus with ground during one gait cycle (**$p<0.01$, *$p<0.05$).

In general, RH did not show any asymmetry in foot contact duration ($p>0.05$). Statistical analyses performed to examine the effect of velocity on contact duration revealed that LC contact duration was longer at 1.5 km/h compared to 3 km/h ($p<0.01$) and 4 km/h ($p<0.01$). RC showed shorter contact duration at 4 km/h compared to 1.5 km/h ($p<0.05$) and 3 km/h ($p<0.005$).

Table 4-2. Stance phase time of both LH and RH, normalised into 100 % of gait cycle (Mean $\pm$ SD). LH showed asymmetry at 4 km/h, while RH showed asymmetry at 1.5 km/h.

	Left-handers stance phase time (in % gait cycle)			Right-handers stance phase time (in % gait cycle)		
Velocity	Left gait cycle	Right gait cycle	*p*	Left gait cycle	Right gait cycle	*p*
1.5 km/h	76.8 $\pm$ 4.9	73.5 $\pm$ 6.9	0.3	76.3 $\pm$ 4.5	72.8 $\pm$ 3.6	0.04*
3 km/h	72.7 $\pm$ 1.7	69.5 $\pm$ 2.1	0.1	72.5 $\pm$ 2.8	71.2 $\pm$ 3.7	0.45
4 km/h	74.2 $\pm$ 7.2	70.2 $\pm$ 6.1	0.04*	73.0 $\pm$ 3.0	69.7 $\pm$ 3.8	0.07

As shown in Table 4-2, among LH, stance phase of left gait cycle was significantly longer than that of right gait cycle in 4 km/h ($p<0.05$). On the contrary, among RH, it was in 1.5 km/h stance phase of left gait cycle was found longer than right gait cycle ($p<0.05$). There was no effect of velocity on stance phase observed in LH ($p>0.05$). In

RH, effect of velocity was observed only in stance phase of left gait cycle where stance phase at 3 km/h was shorter than that at 1.5 km/h ($p<0.05$).

4.3.4 Foot Pressure Cross-Correlation Function

There was no significant differences of both CCF coefficient and time lag of great toe, 3rd metatarsal and 5th metatarsal pressure compared between different velocities as well as between LH and RH ($p>0.05$). As shown in Figure 4-4, all CCF coefficients of 1st metatarsal were found close to 1 which indicates highly symmetrical. There were no significant differences between velocities in the same group as well as between groups ($p>0.05$). Mann-Whitney U-test performed on 3 km/h and 4 km/h of manual pushing found significant differences between both groups ($p<0.05$), where time lags of RH were close to zero, while LH had time lag 3.2 in 3 km/h and 2.3 in 4 km/h, indicating their R1MT touch the ground earlier than their L1MT.

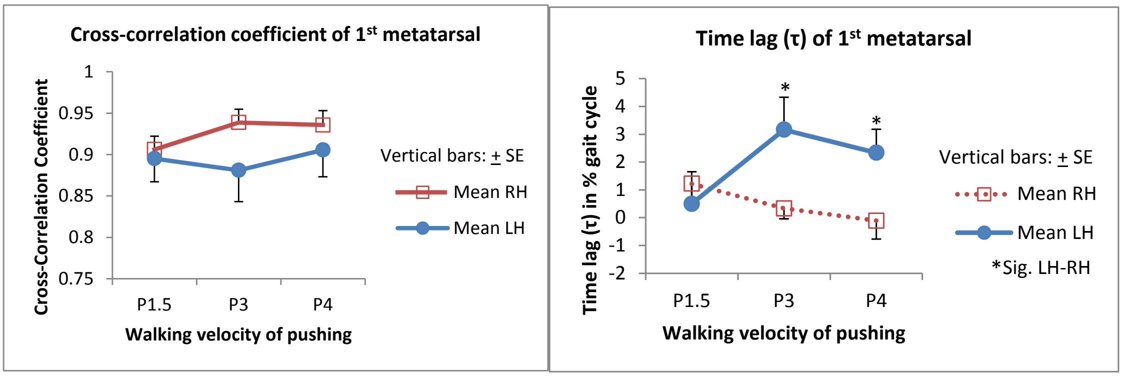


Figure 4-4. CCF coefficient (R_{xy}) and time lag (τ) of 1st metatarsal (mean $\pm$ SE). Time lag of 1st metatarsal showed that LH had greater time lag than RH at 3 and 4 km/h (*$p<0.05$).

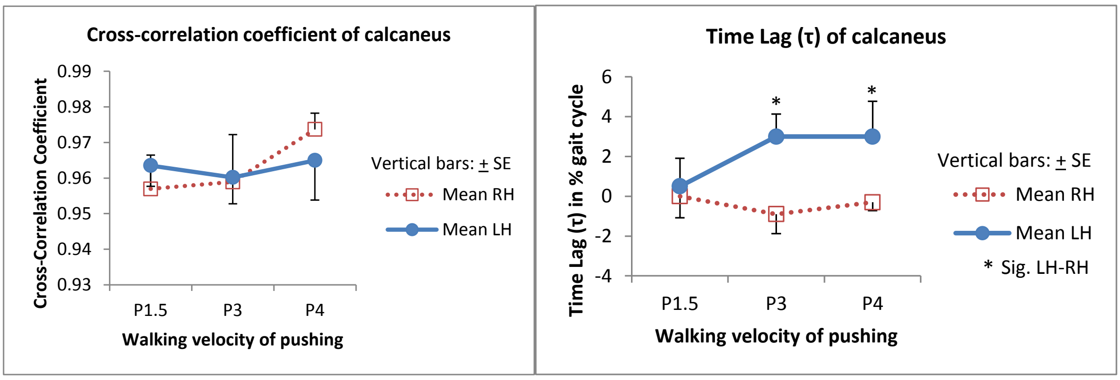


Figure 4-5. CCF coefficient (R_{xy}) and time lag (τ) of calcaneus (mean ± SE). Time lag of calcaneus showed that left-handers had greater time lag than right-handers at 3 and 4 km/h (*$p<0.05$).

As shown in Figure 4-5, highly symmetrical pressure was also showed in calcaneus, where there were insignificant differences between velocities and group of handedness ($p>0.05$). In the same group of subjects, time lag was not influenced by velocity ($p>0.05$). However, in comparison between both groups of handedness, significant differences were observed in 3 km/h ($p<0.05$) and 4 km/h ($p<0.05$), where LH time lag was found to be positive and RH time lag was found to be negative but relatively close to zero.

4.3.5 Pushing Force

As shown in Figure 4-6 (a), pushing force generated by LH at 1.5 and 3 km/h velocities showed three peaks of pushing force that are at the beginning, the middle and at the end gait cycle, representing heel strike, mid-stance, and the end of swing phase. All were occurred during double support. At 4 km/h of manual pushing, pushing force did not show any distinctive peak. RH exerted pushing forces whose patterns did not change between different velocities, where the greatest peak of pushing force during left gait cycle occurred in mid-stance, while the greatest peak of pushing force during right gait cycle occurred during heel-strike and the end of swing phase, both refer to the timing between left foot pre-swing and toe-off (Figure 4-6 (b)).

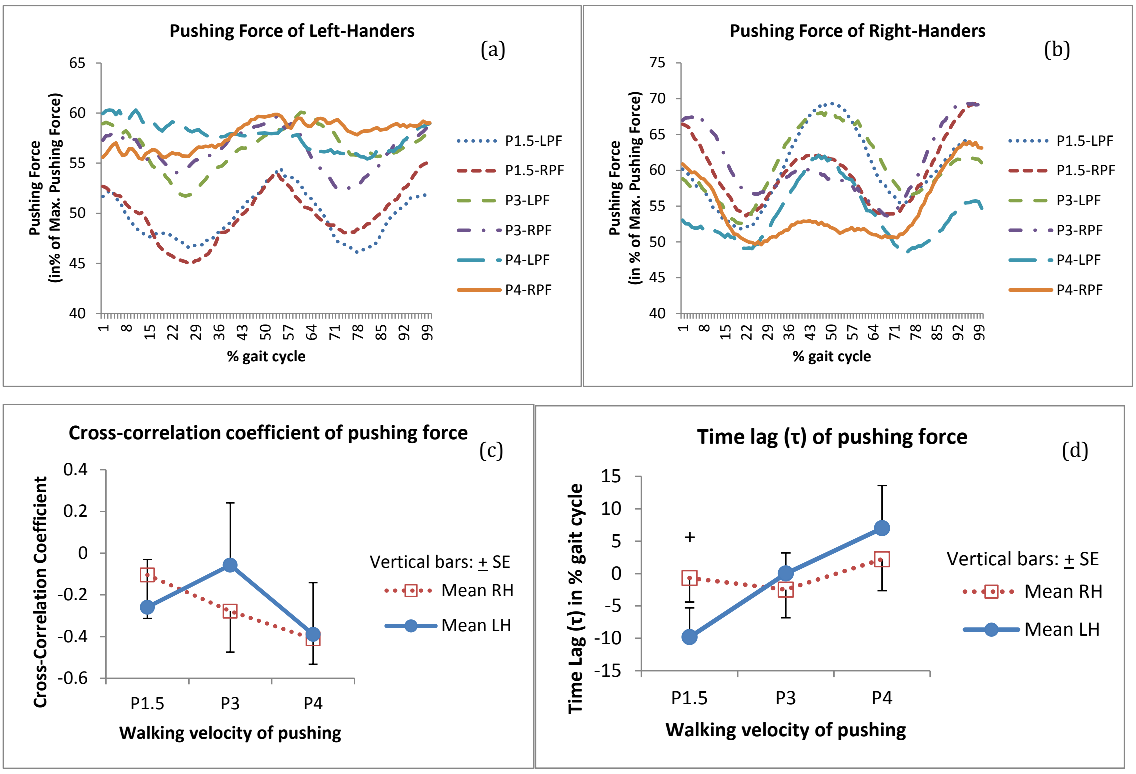


Figure 4-6. Grand average of pushing force of LH (a) and RH (b), where LPF refers to pushing force of left gait cycle and RPF refers to pushing force of right gait cycle. CCF coefficient (R_{xy}) (c) and time lag (τ) (d) of pushing force (mean±SE). There was no significant difference found (+p<0.1).

As displayed in Figure 4-6(c), CCF coefficients are low, close to zero, with slight tendency toward anti-phase as indicated by negative value. However, CCF coefficients and time lags of pushing forces did not find any significant differences between velocities and between LH and RH (p>0.05).

4.3.6 Muscle activation

As shown in Figure 4-7 (a), in TA muscle activation, there were no significant differences of CCF coefficients due to velocity or laterality (p>0.05). Time lag analysis found the effect of velocity on LH, where manual pushing at 1.5 km/h resulted in negative time lag, which is significantly different from 3 and 4 km/h (p<0.05). RH, on the contrary, did not show any significant differences in time lag due to velocity (p>0.05). We also observed significant differences between LH and RH in time lag at 1.5 km/h (p<0.05) and 4 km/h (p<0.05), which were mainly caused by the change in LH.

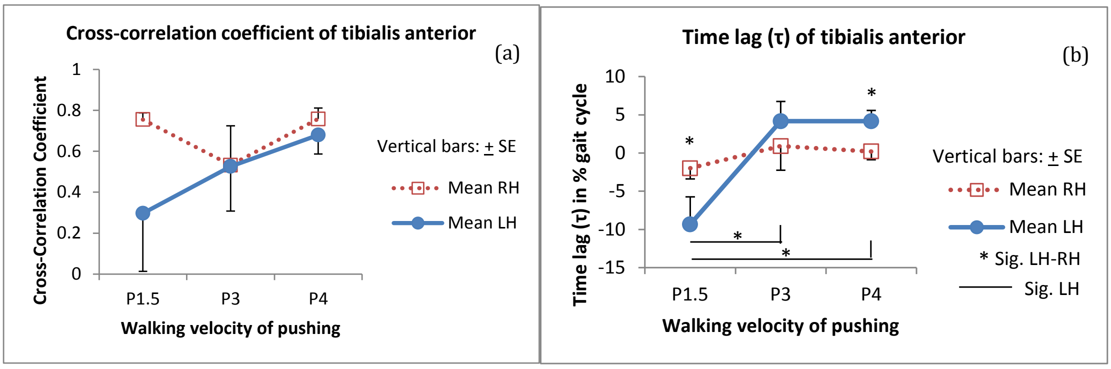


Figure 4-7. CCF coefficient (R_{xy}) (a) and time lag (τ) (b) of TA muscle (mean±SE) (*p<0.05). Time lag of left-handers at 1.5 km/h was significantly different from that at 3 and 4 km/h. At 1.5 and 3 km/h, left-handers showed different time lag from right-handers.

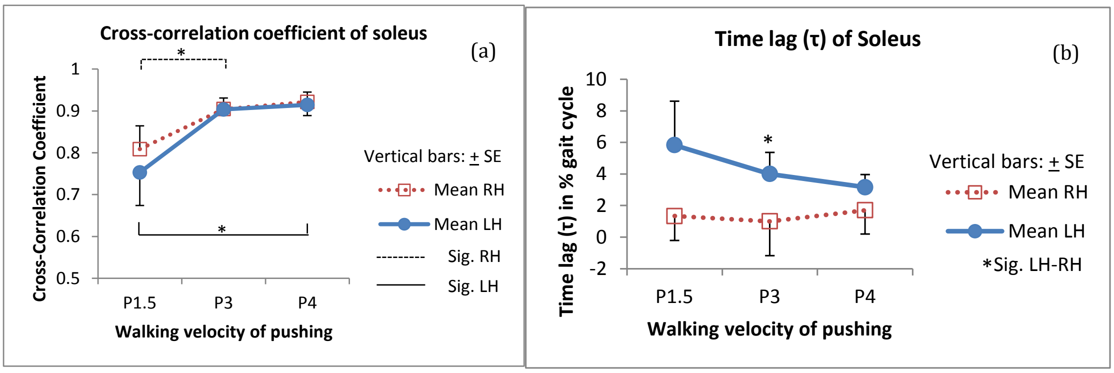


Figure 4-8. CCF coefficient (R_{xy}) (a) and time lag (τ) (b) of Sol muscle (mean±SE) (*p<0.05). Left-handers showed significant difference in CCF coefficients between 1.5 and 4 km/h, while right-handers showed significant difference between 1.5 and 3 km/h. Left-handers and right-handers showed difference between groups of handedness at 3 km/h.

As shown in Figure 4-8(a), Wilcoxon signed-rank test revealed significant difference in LH Sol muscle CCF coefficients between 1.5 and 4 km/h where the coefficients increased from 0.75 to 0.91 (Z=-2.201, p<0.05). RH showed significant difference in CCF coefficients between 1.5 and 3 km/h where the coefficient increased from 0.81 to 0.9 (Z=-2.429, p<0.05). Both LH and RH did not show significant difference in time lag due to velocity change (p>0.05). As shown in Figure 4-8(b), LH showed significantly greater time lag than RH at 3 km/h (p<0.05).

We did not observe any significant differences due to velocity changes in both L5ES muscle CCF coefficients and time lags in both LH and RH (p>0.05). Further statistical analyses in comparison between both groups of handedness also did not find any significant differences (p>0.05).

As shown in Figure 4-9(a), Friedman test found significantly different CCF coefficients in Triceps muscle of LH ($X^2(2)=9.333$, p<0.01). Wilcoxon signed-rank test revealed that significant CCF coefficients differences were in existence between manual pushing at 1.5 km and 4 km/h and between 3 and 4 km/h (p<0.05). RH did not show any significant differences in any statistical analyses performed (p>0.05). Furthermore, both groups of handedness did not show any significant differences in all the three velocities (p>0.05).

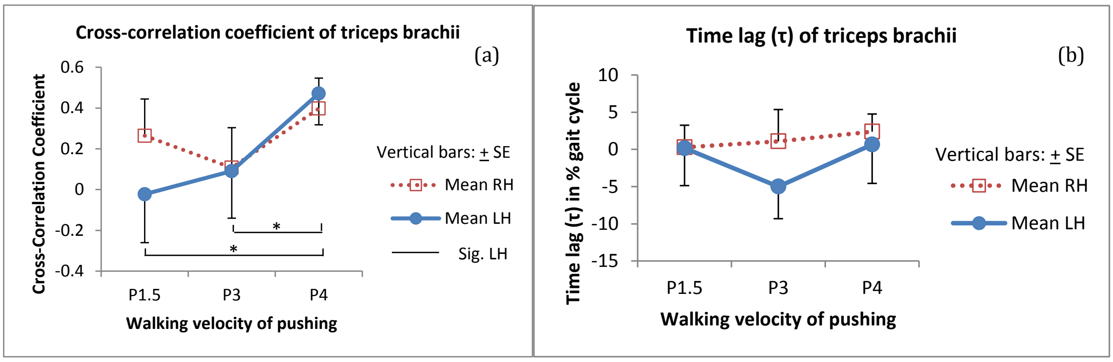


Figure 4-9. CCF coefficient (R_{xy}) (a) and time lag (τ) (b) of Triceps muscle (mean±SE) (*p<0.05). Left-handers showed increase of CCF coefficients between 1.5 and 4 km/h and between 3 and 4 km/h.

4.4 Discussion

4.4.1 Asymmetry in anthropometry data

While some data showed asymmetry in both LH and RH we assume them as normal since subjects did not show any dysfunction in locomotion. Lower acromion height on the right side accompanied with higher iliac crest on the right side have been reported in previous studies among healthy right-handers (Gossman et al., 1982; Kendall and

McCreary, 1983; Neumann et al., 1988) as observed in this study. Kendall and McCreary (1983) associated this phenomenon with a pelvic posture commonly maintained during standing by right-handers when the right hip is in abducted position and left hip is in slightly abducted position, and the elongated condition of the right hip in longer term will result in asymmetrical posture (Kendall and McCreary, 1983; Neumann et al., 1988). Greater trochanter showed asymmetrical height in LH. Static pelvic asymmetry (SPA) was reported in healthy subjects without any sign of dysfunction, and usually caused by asymmetrical load (Gnat and Saulicz, 2008). SPA arises from various sources such as asymmetrical position of sacroiliac joints, asymmetrical muscle tone around the pelvis, leg length discrepancy, and lateral curvatures affecting lumbar spine, which can be eliminated by therapy (Gnat and Saulicz, 2008). Asymmetrical limbs diameter due to laterality is a common phenomenon, and arm circumference has been used to assess handedness since the 19th century (Jones, 1870).

4.4.2 Gait cycle duration is symmetrical

Despite of significant change into shorter gait cycle duration due to increasing velocity, we did not observe asymmetrical gait cycle in both groups of handedness in all velocities measured. Holden et al. (1984) suggested that gait cycle duration difference between left and right foot was unreliable for assessing physical therapy treatment outcomes in subjects with hemiplegia, supported by more recent study (von Schroeder et al., 1995). Further, we should not neglect the effect of treadmill walking which was reported to have greater effect on gait stability compared to walking on a static ground (Waagfjord et al., 1990; Norman et al., 1995; Alton et al., 1998; Laufer et al., 2001). If gait cycle duration is considered unreliable in assessing patients with highly asymmetrical gait patterns, it ought to be more unreliable to be used on healthy subjects. Gait cycle duration only measures foot contact duration globally and tend to neglect difference due to foot shape between humans and other animals.

In the study of locomotion evolution, Stewart and Golubitsky (2011) introduced the concept symmetry breaking in which symmetrical system consists of smaller subgroups of the whole system with various symmetrical characteristics. Therefore, we need to examine further ground contact duration of the five points of footsole measured in order to identify whether there is asymmetry among those points. The humans' foot is characterised with narrow width and longer length, and less stable compared to non-human primates such as chimpanzee which has wider width and abducted great toe (Kiriyama et al., 2005). This foot shape allows faster progression at the expense of medio-lateral stability (Kiriyama et al., 2005), and also suggests that the contact timing of various parts of footsole should vary.

4.4.3 LH showed asymmetry in calcaneus contact duration whereas RH showed symmetrically shortened calcaneus contact duration due to velocity increase

LH showed asymmetrical calcaneus contact duration in all velocities and RC contact duration shortened between 1.5 and 4 km/h. On the other hand, RH showed symmetrical contact duration in all velocities, with both LC and RC had shorter contact duration in higher velocity. During walking at low velocity, all area of the soles of the feet contacts the ground, while at higher velocity of walking, the foot on the front touches the ground with heel, while the foot behind touches the ground with the ball (Cavagna and Margaria, 1966). The condition is represented by foot contact duration in this study.

CCF analysis discovered that while RH had time lag close to zero, RC and R1MT of LH touched the ground earlier than their LC and L1MT at 3 and 4 km/h. Previous study by Cameron and Adams (2003) on kicking found that left-footers utilized their preferred foot predominantly, while right-footers did not show any left-right difference. As kicking requires greater force than walking, we assume that this study probably more comparable to manual pushing. However, while that study employed left and right-footers, our study involved mixed and right-footers. Despite more symmetrical, RH were found to be more sensitive to increasing velocity where bilaterally shortened contact duration on calcaneus showing that RH performed more adaptation than LH. Neuroanatomical-based advantage of LH was suggested as the specialisation of the right hemisphere for spatial processing resulted in advantage in adaptation to task with spatial requirements (Carnahan and Elliott, 1987), where parietal lobe of the right hemisphere controls the perception of location of object and that of left hemisphere for plans for movements (Carlson, 1992).

Based on energetic optimisation during walking, a study which employed bipedal walking models suggested that more symmetrical walking requires less work than asymmetrical walking (Srinivasan, 2011). Therefore, more asymmetrical data of LH compared to RH suggested that while they are less sensitive to velocity change, they probably possess disadvantage in less optimal energetic cost. Muscular activity in walking is aimed to compensate the energy loss at each step (Cavagna et al., 1964), thus LH should show greater alteration in muscles activation.

4.4.4 Pushing force mirrored ground reaction force during walking

We did not find any significant difference in pushing force in any condition measured by CCF coefficient and time lag. During gait cycle, pushing force reached its peak three times, at the beginning, the middle, and the end of gait cycle. These phenomena occurred concurrently with the end of stance phase of each foot especially during double stance. While pushing, subjects try to arrange their posture in order to optimise the chain of force vectors from the ground contacting lower limbs through his body to the point of force application (Kroemer, 1969). In our study, pushing involves walking which is one of the most energetically demanding daily activities (Bertram, and Hasaneini, 2013) and its main objective is to minimise energy expenditure (Cavagna et al., 1977a). There are two strategies proposed: energy recovery within a system based on assumption that bouncing gaits facilitate exchange between energy forms (Cavagna et al., 1963; Cavagna et al., 1964; Cavagna and Margaria, 1966; Cavagna and Kaneko, 1977b), and ground collision-energy loss compensation strategy by performing push-off pre-emptively at the foot of the trailing limb before the heel-strike of the next stance foot which can reduce the loss up to 66% (Kuo, 2002; Bertram, and Hasaneini, 2013).

The vertical ground reaction force during walking has been recognised to have double-humped profile with two force spikes at the beginning and the end of stance phase (Ackermann and van den Bogert, 2010; Srinivasan, 2011). In our study pushing force during stance phase clearly mirrored the double-humped profile. During heel-strike and push-off, centre of mass of the body was situated at its lowest point, representing the high energetic-cost portion which resulted in the spike, whereas low energetic-cost portion where body centre of mass was at its highest point on a single limb support was responsible for the valley of the pushing force. The energetic-loss compensation strategy related to the change from dorsiflexion to plantarflexion during stance phase. This strategy was performed by shortening calcaneus contact duration in higher walking velocity of pushing without reducing ball of foot contact time, since ball of foot is very important during push-off to generate pushing force. Based on statistical analysis, it is on the second strategy the difference between LH and RH exists. While LH only applied the strategy on the right foot, RH applied it on both feet.

4.4.5 Stance phase duration is not shortened due to increasing velocity during manual pushing

Except in left gait cycle of RH between 1.5 and 3 km/h, our study did not find any significant changes due to increasing velocity. Lacquaniti et al. (2012) suggested that stance phase duration during walking has a negative correlation with velocity. The different data between our pushing study and previous study on walking indicate pre-emptive push-off is more crucial in manual pushing than walking, since pushing is a more energy demanding task, energy loss should be minimised by maintaining constant ratio of stance phase to gait cycle duration. This is in agreement to proposition from Pang and Yang (2000) which suggested that stance phase is prolonged and swing phase is delayed when the load on the limb is increased. The findings in this study suggested that laterality probably affected subjects in their strategy to control body movement and the cost of leg work required to perform the movement. In this study we also examined walking with arm-swing condition, where our findings confirm the results of previous studies. Previous study which observed walking on treadmill reported that stance phase duration was reduced from above 75% of gait cycle at 1 km/h to below 65% at 5 km/h (Lacquaniti et al., 2012), while our results found that between 1.5 km/h to 4 km/h of walking with arm-swing, the stance phase duration was reduced from around 74% to 67% with insignificant variation between left and right gait cycle. However we do not report walking condition results in detail in this study, and the data is being prepared for another publication.

4.4.6 Difference between groups of handedness in bilateral muscle activation and whole body coordination

Time lag analysis showed that LH activated their left TA muscle earlier at 1.5 km/h, whereas at both 3 and 4 km/h they activated their right TA muscle earlier. RH, however, did not demonstrate any significant difference due to velocity. LH showed more dramatic change in TA muscle activation symmetry due to dorsiflexion and inversion based on significant difference observed in 1st metatarsal and calcaneus. TA muscle has a role in dorsiflexion and assisting in inversion of the foot (Kendall and McCreary, 1983). Previous studies suggested that against the traditional assumption that lower limb muscles are largely controlled by brainstem and spinal mechanism, transcortical pathway also contributes to stretch reflexes in TA muscle which is related to dorsiflexion, supported by other study (Nielsen, et al., 1997, Petersen, et al., 1998).

Sol muscle has a role in plantar flexion of ankle joint (Kendall and McCreary, 1983). LH showed increase of CCF coefficients in Sol muscle activation from 1.5 to 4 km/h, while RH showed it from 1.5 to 3 km/h, which imply that in greater velocity the temporal waveform become more symmetrical between LSol and RSol muscle. We speculate that this is related to inhibition to direct cortical-motoneuronal coordination by CPGs which is associated to shorter latency of muscle activation and control of rhythmic interlimb coordination (Dietz, 2002) at higher gait velocity. We also observed that at 3 km/h, LH showed significantly higher time lag than RH. Other than LH and RH difference explained previously, this is probably also related to recovery curve of Hoffmann reflex of soleus muscle which was reported to be asymmetrical and influenced by laterality (Goode et al., 1980; Tan, 1985). Sol muscle is an antagonist of TA muscle (Edamura et al., 1991), however, we found that TA and Sol muscle adaptation was rather complementary in coordination. In general, increase of velocity affected time lag of TA muscle in LH and CCF coefficients of Sol muscle in LH and RH. Furthermore, TA muscle time lag showed difference between groups of handedness at 1.5 and 4 km/h whereas Sol muscle time lag showed difference at 3 km/h. More asymmetrical lower limbs muscular activation in LH probably indicates greater degree of direct cortical-motoneuronal control than caudal CPG control.

CCF coefficient and time lag of L5ES muscle did not show any significant difference in both LH and RH or between groups of handedness indicating that different from limbs muscles, L5ES is generally symmetrical, and did not change significantly across all conditions. In the plane of progression, during gait, L5ES muscle activation and almost equal and opposite hip extensor moments are generated to prevent unbalancing of the trunk early in the stance phase (Kavanagh, 2009). The symmetry of L5ES for stabilizing act is not affected by laterality and velocity during manual pushing.

LH showed significant increase in CCF coefficients of triceps muscle activation between 1.5 and 4 km/h and between 3 and 4 km/h, suggesting more symmetrical trend. Such an increase was not observed in RH. Triceps has role in extending elbow joint and additionally assist in adduction and extension of the shoulder joint (Kendall and McCreary, 1983). During manual pushing, rhythmic arm movements are absent, thus we can assume the inexistence of rostral spinal cord CPGs role. The leg cadence was reported to modulate arm cadence (Sakamoto et al., 2007). In other study, small leg displacement was reported to affect Triceps muscle greater than biceps brachii both ipsilaterally and contralaterally (Dietz, et al., 2001). Despite Sakamoto et al. (2006) reported little influence of legs movement on arm cutaneous reflexes, all of these studies agreed on the existence of a task-dependent, flexible neuronal coupling between lower and upper limb muscles. Furthermore, pushing is a cross-symmetrical force exertion where left foot propulsion is translated into right hand pushing force exertion and vice versa, thus LH with mixed-footedness will likely generate more asymmetry than RH with right-footedness.

In general, this study found differences between groups of handedness examined in various variables measured but not in mirrored characteristics between LH and RH. There are many definitions proposed about handedness such as complementary role differentiation (Uomini, 2009) and dynamic-dominance hypothesis (Sainburg, 2002). With regard to footedness, propulsion-control hypothesis was commonly acknowledged (Hirokawa; 1989; Sadeghi et al., 2000). Other study, however, proposed that footedness definition is very task-dependent (Hart and Gabbard, 1997). All of these previous studies reject the assumption that one side is more dominant than the other side to describe laterality, and more inclined to adopt the description as complementary role between sides. Our study, as a study based on whole body coordination task of manual pushing found more about different body coordination patterns between groups of handedness instead of very strict limitation between dominant and non-dominant sides as commonly assumed in the study of laterality. Furthermore, manual pushing involves whole body coordination and cross-symmetrical force exertion, therefore diminishes separation between body sides. Considering that laterality is very task-dependent, further research such as to examine hazardous aspects between right-handers and left-handers in manual pushing or study on manual pushing for therapeutic purposes and so on will require a more controlled experimental procedure.

Relationship between limb laterality and bilateral difference of the brain is highly interesting subject for the researchers in the field of kinesiology and sport sciences, especially in the discussion on direct motor neuronal and intraspinal mechanism role during locomotion, which has been assumed to be very important during evolution (Golubitsky et al., 1999; Dietz, 2002). Asymmetries of cortical organization have been associated with laterality (Springer and Deutsch, 1997; Soros et al., 1999; Jung et al., 2003), with less clear patterns among LH (Coren, 1992; Soros et al., 1999) which is associated with their better ability in utilizing both sides of their limbs due to environmental adaptation as well as lower interhemispheric transfer time and larger corpus callosum (Witelson, 1985, Bernard, 2011). While this study shows difference between groups of handedness, we did not measure brain activities therefore we cannot analyse the relation between brain and asymmetrical locomotion in detail. Laterality itself is affected by individual difference due to cultural and social backgrounds, and in addition, laterality is not necessarily corresponds if all movements of extremities satisfy the regulation of questionnaire which was employed to assess the degree of laterality in this study. In the future, measurement of brain activities using instruments such as electroencephalogram (EEG) or functional near-infrared spectroscopy (f-NIRS) is necessary, to be able to understand better the phenomenon of brain association with laterality.

Other limitation of this study is number of subjects that consisted of only two groups of handedness caused this study unable to go into detail of relation of handedness and footedness. To be able to do such a study, ideally we need to collect data from four combinations of laterality: right-handed and right-footed, left-handed and right-footed, right-handed and left-footed, and left-handed and left-footed subjects. The greater number of subjects from each group based on statistical power analysis is also very crucial for future studies, especially non-right-lateralised subjects which in our study and also in most previous studies were available in only small number. A study involving all those groups will probably able to study further on the degree of brain and CPGs control on locomotion between different groups of laterality.

4.5 Conclusions

In summary, in a whole body coordination movement such as manual pushing, laterality is indicated in a very different pattern from what generally assumed in unimanual or bimanual tasks used to examine handedness as well as unipedal or bipedal tasks to examine footedness. This study shows that left-handers control their whole body movement differently from right-handers, thus they have different musculoskeletal hazard than the majority of the population, despite the symmetrical nature of manual pushing task. In whole body coordination, the definition of more skilled dominant side to describe the preferred side should be avoided, since we could not separate one side from the other. As an initial study, this study very limited to find general characteristics showed by different group of handedness during manual pushing, and unable to assess deeper to every aspect measured. Comprehensive investigation on laterality during whole body movement, thus, require further studies which is more detail in human body parts measured as well as more sophisticated experiment procedure. Future study on association of brain and limb laterality by measuring brain activities involving sufficient number of subjects is necessary.

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Chapter 5

The influence of laterality and walking speed on walking symmetry during gait cycle

5.1 Introduction

Symmetry is a phenomenon which is very common in the universe. In the study of symmetry, there is a paradoxical phenomenon called as symmetry breaking where the symmetric system starts to behave less symmetrically, when the symmetry of the resulting state of the system is a subgroup of the symmetry group of the whole system (Stewart and Golubitsky, 2011). During walking, both legs move half a period out of phase which shows symmetry breaking. In quadrupeds two types of symmetry are observable, namely spatial symmetry which shows the interchanging of fore and hind legs and spatio-temporal symmetry which shows the interchanging of left and right legs with a half-period phase shift, whereas in bipeds, only the latter is observable (Golubitsky et al., 1999). In measuring gait, two important aspects are commonly used, namely gait cycle which represents the time duration between foot-strike of the same leg and stance phase which represents the time when the foot touches the ground (Stewart and Golubitsky, 2011).

One representative of symmetry measure is laterality. Laterality refers to asymmetrical preferential use of limbs and sensory: handedness, footedness, eyedness, and earedness (Schneiders, et al. 2010). In evolutionary studies, bipedalism has been associated with the development of laterality. Study on chimpanzees found that a bipedal stance without hand support will evoke hand preferences (Braccini, et al. 2010). Study on Sichuan snub-nosed monkeys also found that foot preference was significantly stronger in bipedal action (Zhao, et al. 2008). About 85% of human population are right-handed (Uomini, 2006). Other suggested around 90% of humans are right handed, around 80% are right footed and around 70% are right eyed (Carey, et al. 2001). Based on population studies, right foot tends to be the preferred foot of right-handers, however the situation remains unclear for left-handers (Chibber and Singh, 1970; Chapman, et al. 1987; Peters, 1988).

Left-footers were found to control their unipedal posture differently from right-footers and footedness influences postural control which result in differences in balance perception and action (Golomer and Mbongo, 2004). However, Hart and Gabbard (1997) suggested that lower limb choice for postural stabilization in bipedal context may be independent of that in unipedal, where dominant leg is used for more demanding task. Other study reported that there was no lateral dominance during static single-leg stance, however higher frequency of movement strengthens lateral dominance for postural stability (Kiyota and Fujiwara, 2014). A study on footedness during walking using footprint method found that footedness does not affect gait parameters (Zverev, 2006). In study of gait initiation, Hesse et al. (1997) found that in normal subjects gait initiation is highly symmetrical. This suggestion was negated by research review from Sadeghi, et al. (2000) who hypothesized asymmetrical lower limbs behaviour during gait as a reflection of natural functional differences in propulsion and control. This was supported by Dessery et al (2011) who observed asymmetrical frontal body motion influenced by footedness in gait initiation. Herzog, et al. (1989) measured normal human gait using symmetry index found that gait asymmetries were larger than expected.

Stewart and Golubitsky (2011) hypothesized that the faster the movement, the more a system is stressed, the less symmetry it will become. This is supported by study which found greater propulsion on dominant limb compared to non-dominant limb during fast walking (Seeley et al., 2008). On the other hand, better limbs coordination to recover from perturbation was observed in faster walking but not associated with gait stability (Krasovsky et al., 2014). Furthermore, Lythgo et al. (2011) compared gait symmetry between primary school-aged children and young adults did not observe effects of speed on symmetry. In general both symmetry and asymmetry are observed during gait, however they cannot be generalized in association with lateral dominance (Gundersen et al., 1989).

We assumed that these mixed results from previous studies are actually consistent with symmetry breaking concept (Stewart and Golubitsky, 2011) where in dynamic systems, when symmetry becomes less stable and turns into asymmetry, there will be a reaction to recover the symmetry, thus both conditions are observed. Furthermore, symmetry breaking includes subgroups of symmetrical system, therefore an analysis into smaller units such as measurement of various points of footsole instead of global ground reaction force will be able to describe more detailed symmetry during gait. This study investigated the influence of laterality and walking speed on gait spatio-temporal symmetry by measuring muscular activation and foot pressure.

5.2 Methods

5.2.1 Subjects

Subjects were 17 healthy young adult males obtained from Chiba University student population (age 28 $\pm$ 5 years; height 169.9 $\pm$ 6.9 cm; weight 64.6 $\pm$ 7.3 kg). The laterality data of subjects were measured by Waterloo Footedness Questionnaire (WFQ) and Waterloo Handedness Questionnaire (WHQ) (Elias, et al. 1998), where 11 subjects were found to be both right-handed (50.25 $\pm$ 12.42) and right-footed (9.25 $\pm$ 4.13), and 6 subjects were grouped into left-handers (-15.17 $\pm$ 13.7) and mixed-footers (0.0 $\pm$ 7.5). To ease grouping we call the first group as right-handers (RH) and the second groups as left-handers (LH). Prior to the experiment, subjects were given informed consent which was approved by the Ethics Committee of the Graduate School of Engineering, Chiba University (25-25).

5.2.2 Instruments

Electromyogram (EMG) data were collected bilaterally from four muscles: tibialis anterior (TA), soleus (Sol), lumbar erector spinae (ES) and triceps brachii (TB). The strain amplifier output and EMG electrodes were connected to a Biopac MP 150 data acquisition system (Biopac Systems, USA), which then was connected to a personal computer. Ten FSR-400 pressure sensors (Interlink Electronics, USA) were attached bilaterally on great toe (T; right great toe-RT; left great toe-LT)), 1st metatarsal (1MT; left 1st metatarsal-L1MT; right 1st metatarsal-R1MT), 3rd metatarsal (3MT; left 3rd metatarsal-L3MT; right 3rd metatarsal-R3MT), 5th metatarsal (5MT; left 5th metatarsal-L5MT; right 5th metatarsal-R5MT), and calcaneus (C; left calcaneus-LC; right calcaneus-RC) adapted from a method used by Kiriyama, et al. (2005) in order to measure foot pressure and contact duration. All subjects wore the same footwear to standardise friction. Each trial period was determined by a visual display timer and light sensor (Kodenshi Corp., Japan).

5.2.3 Experiment Procedure

Subjects walked on a treadmill (SportsArt Fitness, Taiwan) in three speeds: 1.5 km/h (W1.5), 3 km/h (W3), and 4 km/h (W4). Before the main trials, subjects were familiarised with treadmill walking. Each trial was performed in 1 minute of which 10 seconds most consistent middle part was recorded. The trial was performed three times. A three minutes rest was given between trials. During trial, subjects were instructed to gaze at the monitor which display timer. Gazing on a monitor is advantageous, because it prevents asymmetrical visual influence on balance control as reported in previous studies (Bessou, et al., 1999; Golomer and Mbongo, 2004; Nagano, et al., 2006).

5.2.4 Data Analysis

Data analysis was based on one gait cycle by using foot pressure sensor as reference. We divided data into right and left side gait cycle, where data from left side muscles and pressure sensor were analysed according to left foot gait cycle, and data from the right side were analysed according to right foot gait cycle. EMG and pressure sensor data were collected at a 1000-Hz sampling rate. Raw EMG signals were band-pass filtered between 15-250Hz, and root mean square (RMS) was derived. All data were normalized into 100 data points in one gait cycle. From the 10 seconds recorded data, 3 gait cycles were taken into analysis.

To measure spatio-temporal change in symmetry of variables measured in left and right side, foot contact duration and cross-correlation function (CCF) was employed. The coefficients (CCF_{coeff}) vary between -1 and +1, where a positive correlation value indicates the signals are in phase, and a negative value indicates inverse relationship (Nelson-Wong et al. 2009). CCF also measured time lag (τ) between the two signals. Data measured from the left and right sides were arranged so that if the peak of right side were earlier than the left side, time lag will be positive, while if it were later, time lag will be negative.

In statistical analyses, subjects were grouped into right-handers (RH) and left-handers (LH) groups. All recorded data as well as CCF_{coeff} data were averaged between subjects in the same group. We analysed the characteristics of each group, and then for CCF_{coeff} and time lag data we also performed comparison between groups. The Shapiro-Wilk test was used to determine whether parametric or non-parametric method should be used. We employed Student`s paired T-test for parametric method and Wilcoxon signed-rank test for non-parametric method in order to compare data between left and right side. Data comparison between different velocities in the same group was performed using one-way repeated measures ANOVA with Bonferroni post-hoc test for parametric method and Friedman Test with the Wilcoxon signed-rank post-hoc test for non-parametric method. Because the two groups of subjects have unequal samples, in order to compare both groups, independent T-test for parametric method and Mann-Whitney U test for non-parametric method were used. Statistical significance was set at $p<0.05$. All statistical analyses were performed with Microsoft Excel 2010J and IBM SPSS 17J.

5.3 Results

5.3.1 Gait Cycle Duration

Gait cycle duration was found to be shorter with faster walking in both left and right gait cycle of both groups ($p<0.05$) (Table 5-1). However, asymmetry in gait cycle duration in both groups was not observed since there was no difference between left and right gait cycle duration ($p>0.05$).

Table 5-1. Gait cycle duration of both left-handers and right-handers (mean $\pm$ SD). There was no asymmetry observed on gait cycle duration.

	LH gait cycle duration (in second)			RH gait cycle time (in second)		
Speed	Left gait cycle	Right gait cycle	*p*	Left gait cycle	Right gait cycle	*p*
1.5 km/h	1.49 $\pm$ 0.21s	1.49 $\pm$ 0.20s	ns	1.60 $\pm$ 0.38s	1.60 $\pm$ 0.39s	ns
3 km/h	1.19 $\pm$ 0.12s	1.19 $\pm$ 0.12s	ns	1.12 $\pm$ 0.11s	1.12 $\pm$ 0.12s	ns
4 km/h	1.05 $\pm$ 0.07s	1.05 $\pm$ 0.06s	ns	1.00 $\pm$ 0.12s	1.00 $\pm$ 0.12s	ns

5.3.2 Foot Contact Duration

As shown in table 5-2, asymmetrical stance phase duration was not observed both in LH and RH. Both in LH and RH, increase in walking speed from 1.5 km/h to 4 km/h shortened stance phase duration significantly in both feet ($p<0.05$). LH showed significantly shorter stance phase duration on left foot when walking speed increased from 1.5 km/h to 3 km/h ($p=0.02$). RH showed significantly shorter stance phase duration on left foot when walking speed increased from 1.5 km/h to 3 km/h (p=0.01) and from 3 km/h to 4 km/h ($p=0.005$), and on the right foot when walking speed increased from 1.5 km/h to 3 km/h ($p=0.01$).

Table 5-2. Stance phase time of both left and right-handers, normalised into 100 % of gait cycle (mean ± SD).

Stance Phase Time	Left-handers			Right-handers		
Speed	Left gait cycle	Right gait cycle	*p*	Left gait cycle	Right gait cycle	*p*
1.5 km/h	77.0 ± 2.5	74.5 ± 3.0	ns	74.8 ± 1.9	73.6 ± 3.2	ns
3 km/h	73.0 ± 1.2	71.0 ± 2.9	ns	71.5 ± 2.9	70.5 ± 3.2	ns
4 km/h	71.0 ± 2.2	68.5 ± 3.3	ns	69.1 ± 2.9	67.3 ± 2.7	ns

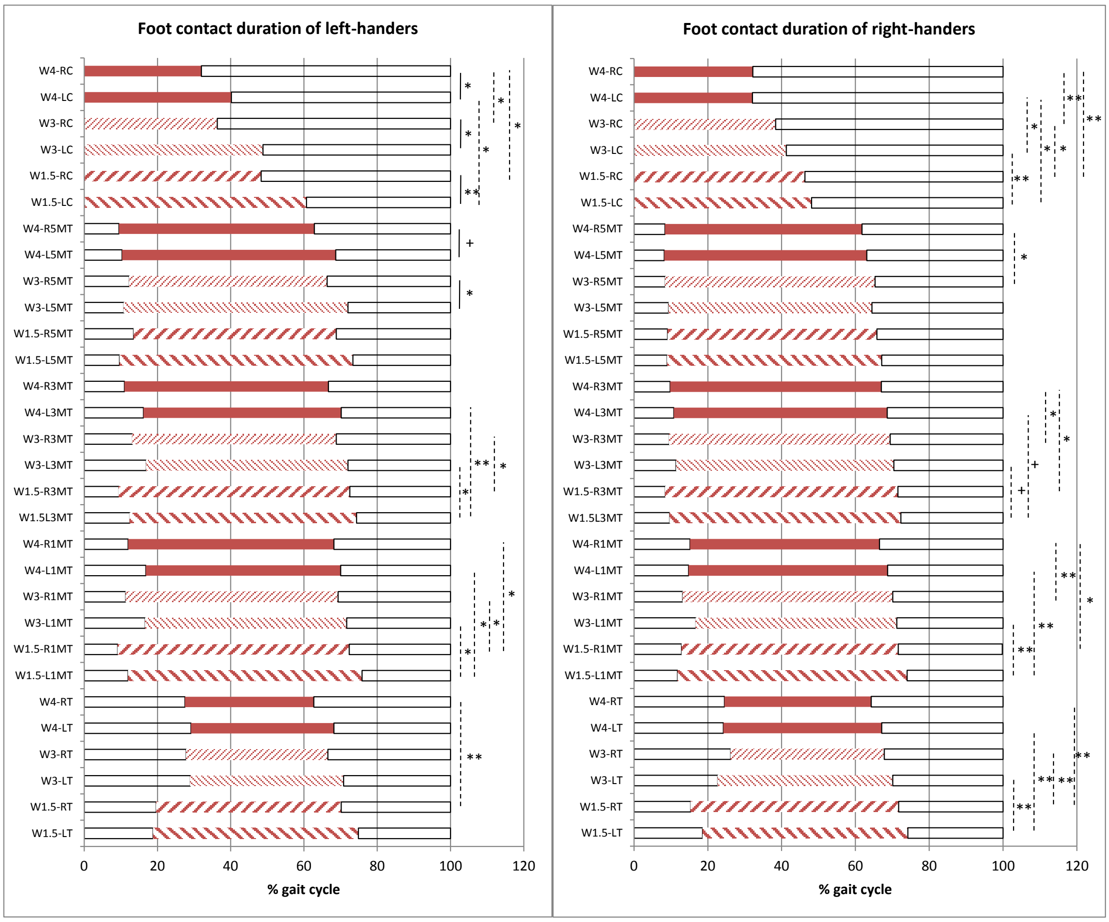

Figure 5-1. Foot contact duration of footsole points ($^+p<0.1$; $*p<0.05$; $**p<0.001$). Straight line shows significant differences between left and right foot. Dashed line shows significant differences between different speeds.

Asymmetrical foot contact duration only showed by LH, especially in C at all speeds ($p<0.05$), and in 5MT at 3 km/h walking speed ($p<0.05$) (Figure 5-1). The contact duration of 5MT was largely unaffected by walking speed ($p>0.05$) except in the right foot of RH between 3 km/h and 4 km/h walking speeds ($p=0.02$). Walking speeds also affected T contact duration in RH than LH where shortened duration was observed between 1.5 km/h and 4 km/h walking speed on the right foot (p=0.007).

5.3.3 Foot Pressure Cross-Correlation Function

As shown in Figure 5-2, LH showed significant increase in CCF_{coeff} of T when walking speed increased from 1.5 km/h to 4 km/h (from 0.87 $\pm$ 0.06 to 0.93 $\pm$ 0.01; p=0.046). RH showed significantly higher CCF_{coeff} of T than LH at 1.5 km/h (0.93 $\pm$ 0.07 vs 0.87 $\pm$ 0.06; p=0.026) and 3 km/h (0.94 $\pm$ 0.05 vs 0.87 $\pm$ 0.06; p=0.039). LH showed significantly higher time lag of T than RH at 3 km/h walking speed (p=0.032).

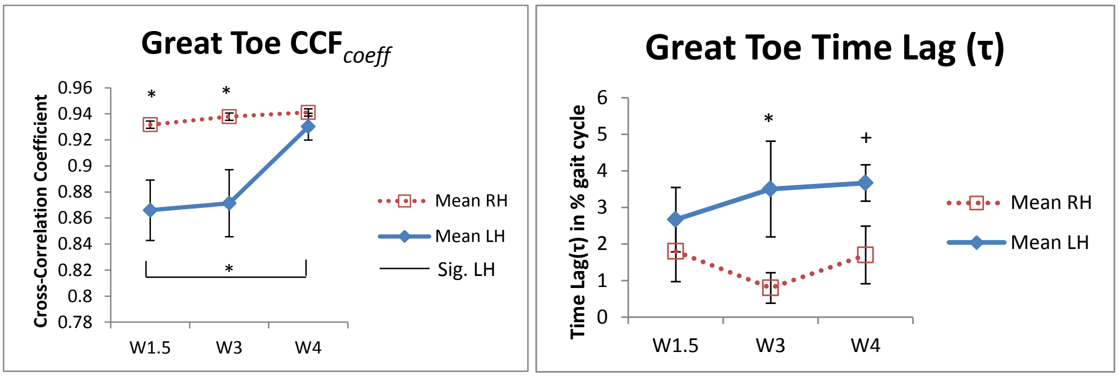

Figure 5-2. CCF_{coeff} and time lag (τ) of great toe (mean $\pm$ SE) ($^{+}p<0.1$; $^{*}p<0.05$). Compared to LH, RH showed significantly greater CCF_{coeff} at 1.5 and 3 km/h, and shorter τ at 3km/h.

Figure 5-3 shows that RH had significantly greater CCF_{coeff} of 1MT than LH at 3 km/h (0.95 $\pm$ 0.03 vs 0.92 $\pm$ 0.02; p=0.02). There were no significant differences observed in time lag of 1MT between groups of handedness and between speeds (p>0.05).

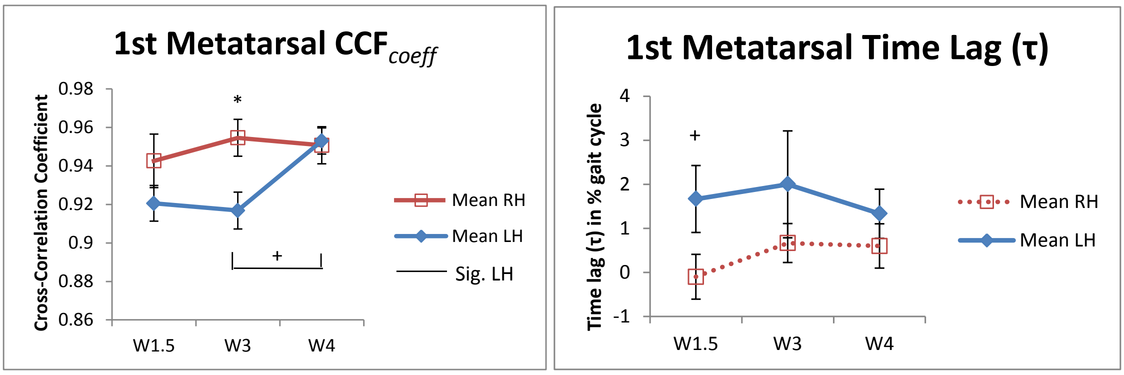

Figure 5-3. CCF_{coeff} and time lag (τ) of 1st metatarsal (mean ± SE) ($^{+}p<0.1$; $*p<0.05$). RH showed significantly greater CCF_{coeff} than LH at 3 km/h.

Figure 5-4 shows that there were no significant differences observed between groups of handedness and walking speeds in 3MT CCF_{coeff} ($p>0.05$). LH showed significantly longer time lag than RH in time lag of 3MT at 1.5 km/h (1.17 ± 1.33 vs -0.20 ± 1.03; $p=0.048$) and 3 km/h (3.00 ± 2.28 vs 0.70 ± 1.16; $p=0.022$).

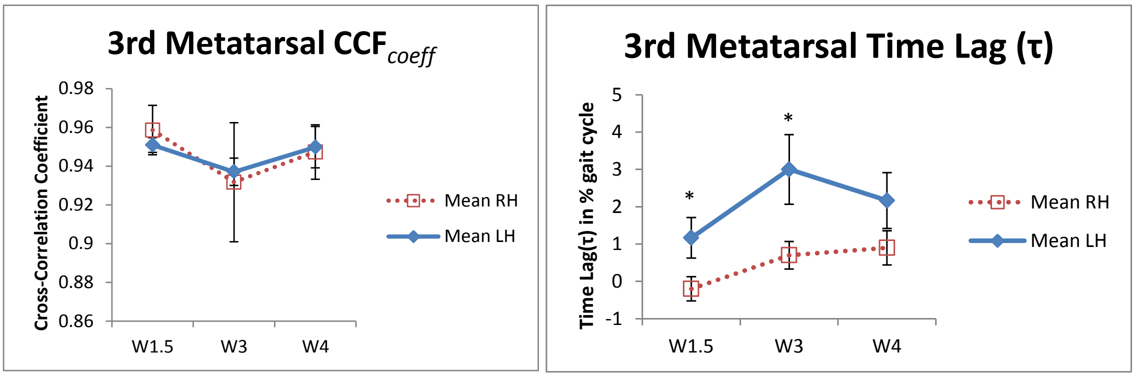

Figure 5-4. CCF_{coeff} and time lag (τ) of 3rd metatarsal (mean ± SE) ($*p<0.05$). LH showed longer time lag (τ) than RH at 1.5 and 3 km/h.

There were no significant differences observed in CCF_{coeff} and time lags of 5MT between groups of handedness and between walking speeds ($p>0.05$), as shown in Figure 5-5. However RH showed tendency of more symmetrical walking as shown by tendency of higher CCF_{coeff} at 3 km/h and 4 km/h as well as shorter time lags at both speeds compared to LH ($p<0.1$).

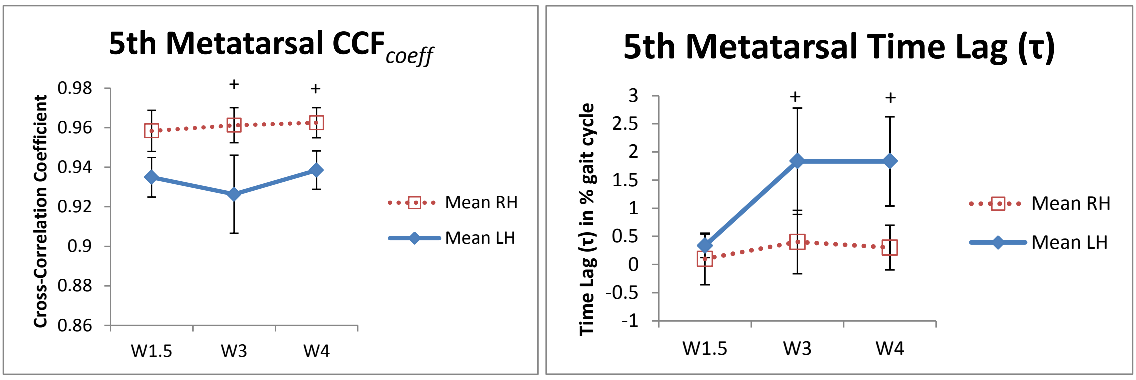

Figure 5-5. CCF_{coeff} and time lag (τ) of 5th metatarsal (mean ± SE) ($^{+}p<0.1$).

There were no significant differences observed between groups of handedness and walking speeds in C CCF_{coeff} ($p>0.05$). As shown in Figure 5-6, LH showed significantly longer time lags than RH at 1.5 km/h (3.67±2.88 vs 0.70±2.45; $p=0.045$), 3 km/h (4.17±2.99 vs 1.20±2.25; $p=0.040$) and 4 km/h (3.17±2.86 vs 0.50±2.01; $p=0.045$).

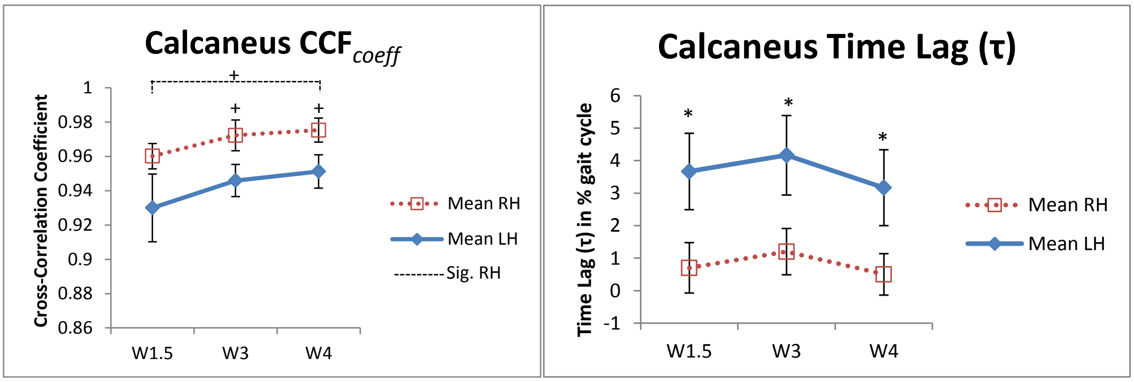

Figure 5-6. CCF_{coeff} and time lag (τ) of calcaneus (mean ± SE) ($^{+}p<0.1$; $^{*}p<0.05$). LH showed significantly longer time (τ) lag than RH at 1.5 km/h, 3 km/h and 4 km/h.

5.3.4 Muscle activation

Figure 5-7 shows that TA muscle activation did not show any significant differences due to laterality and walking speeds in both CCF_{coeff} and time lags ($p>0.05$). RH only showed tendency of greater CCF_{coeff} when walking speed increased from 1.5 km/h to 4 km/h ($p=0.074$).

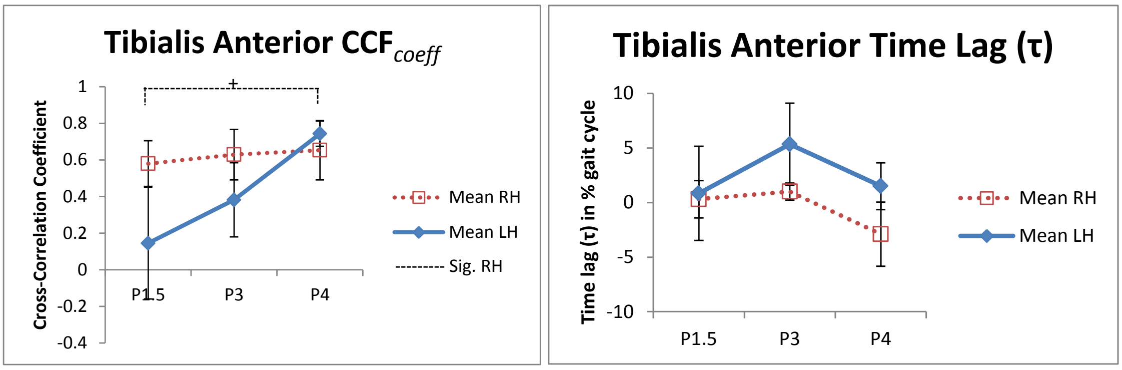

Figure 5-7. CCF_{coeff} and time lag (τ) of tibialis anterior muscle (mean ± SE) ($^+p<0.1$).

Figure 5-8 shows that faster walking speed from 1.5 km/h to 4 km/h resulted in significant increase of Sol muscle activation CCF_{coeff} in RH (from 0.90±0.06 to 0.95 ± 0.02; p=0.011), but not in LH (p>0.05). LH showed longer time lag of Sol muscle activation than RH at 3 km/h walking speed (3.83 ± 2.64 vs 0.60 ± 2.41; p=0.025).

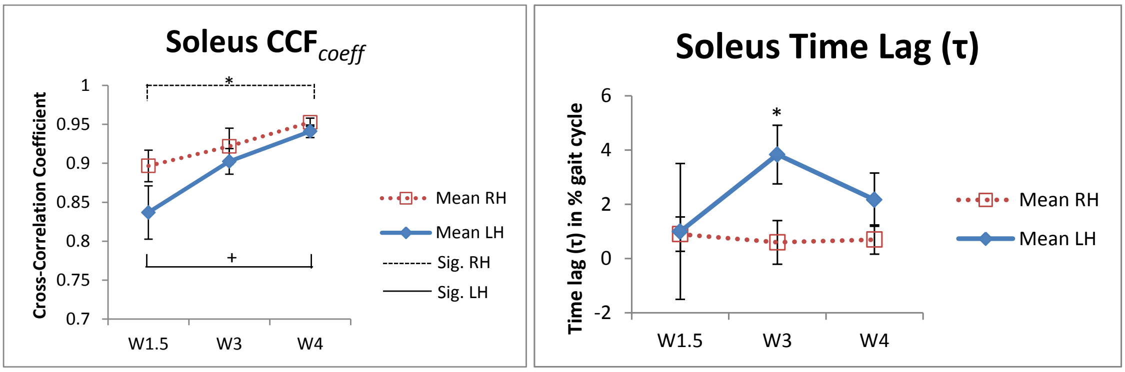

Figure 5-8. CCF_{coeff} and time lag (τ) of soleus muscle (mean ± SE) ($^+p<0.1$; $^*p<0.05$).

As shown in Figure 5-9, RH showed significant increase of ES muscle activation CCF_{coeff} from 1.5 km/h to 4 km/h walking speed (from 0.78 ± 0.11 to 0.91 ± 0.02; p=0.007). This increase also resulted in significant difference between LH and RH in ES muscle activation CCF_{coeff} at 4 km/h walking speed (0.78 ± 0.10 vs 0.91 ± 0.02; p=0.005). Significant time lags differences were not observed in all conditions measured (p>0.05).

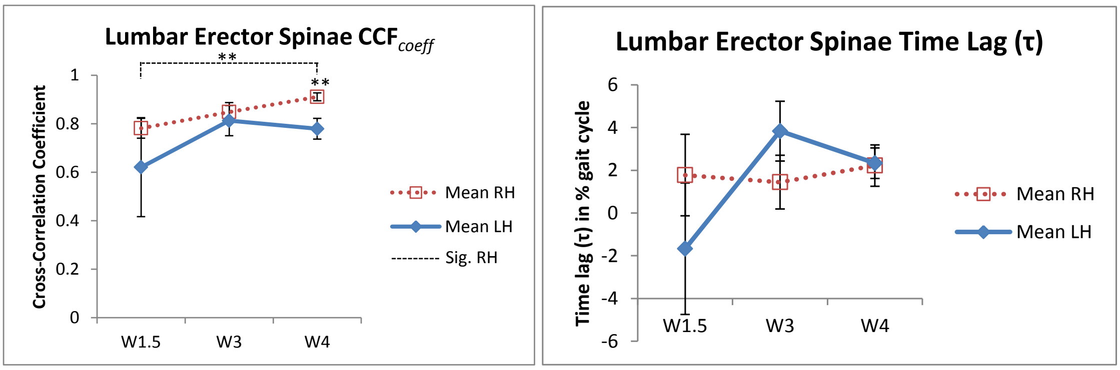

Figure 5-9. CCF$_{coeff}$ and time lag (τ) of lumbar erector spinae muscle (mean $\pm$ SE) (**p<0.01).

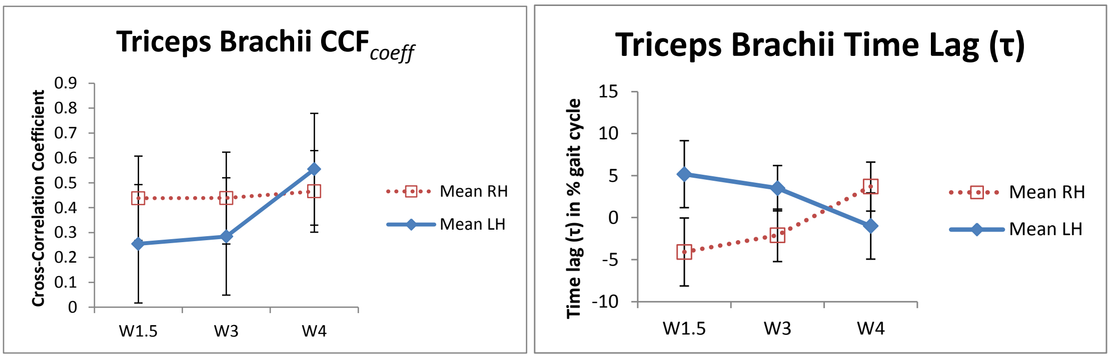

Figure 5-10. CCF$_{coeff}$ and time lag (τ) of triceps brachii muscle (mean $\pm$ SE).

TB muscle did not show any significant difference in both CCF$_{coeff}$ and time lags due to either laterality or walking speeds (p>0.05).

5.4 Discussion

Gait cycle duration was found to be symmetrical which support a study by Golubitsky et al. (1999) on symmetrical gaits in various animals based of simple fractions of lateral phase shifts. In previous studies, gait cycle duration was also found to be symmetrical in post-therapy hemiplegic subjects (Holden et al., 1984), supported by a more recent study (von Schroeder et al., 1995). Step time which is related to gait cycle duration was also reported to be less reliable to identify healthy gait (Owings and Grabiner, 2004).

Stance phase duration was found shorter with faster walking. These data are in agreement with previous studies which observed walking on treadmill, where stance phase duration was reduced from above 75% of gait cycle at 1 km/h to below 65% at 5 km/h (Ivanenko et al., 2004; Lacquaniti et al., 2012). Duty factor of walking is always greater than 0.5 which means stance phase is always longer than 50% of gait cycle, whereas duty factor of running is less than 0.5 (Stewart and Golubitsky, 2011). However significant changes between various walking speeds are not uniform, especially between 1.5 and 3 km/h and between 3 km/h and 4 km/h, thus asymmetry probably exists in the smaller components of footsole or muscle activation, the features retained in long evolutionary process which unlike animal, are not observable without the use of special instruments.

LH showed that their T ground contact duration was only affected by walking speed only on the right foot between 1.5 km/h and 4 km/h, compared to significant effect on both T observed on RH. This asymmetrical effect was supported by CCF_{coeff} which showed significant increase between 1.5 km/h and 4 km/h was found among LH but not in RH. LH also showed greater time lag on T compared to RH at 3 km/h. In 1932, Lambrinudi described the main function of T is to support the metatarsal heads to take the full body weight when the heel is raised (Lambrinudi, 1932; Hughes et al. 1990). In RH, T contact duration between 1.5 km/h and 4 km/h was reduced on both feet, whereas in LH, the LT contact duration was not reduced, thus it gave greater support to metatarsals head of the left foot. The range of motion of the T lies in dorsiflexion which occurs twice during stance phase of walking, namely active and passive dorsiflexion (Bojsen-Moeller and Lamoreux, 1979). In this study, foot pressures sensor only measured passive T dorsiflexion which occurs after the heel leaves the floor prior to push-off, as the toes are forced dorsally by body weight. Other than supporting body weight bearing, T dorsiflexion was also hypothesized to have an effect on venous flow in the metatarsals head of foot (Bojsen-Moeller and Lamoreux, 1979).

In static standing, a study with large number of subjects from preschool children reported slightly greater anterior foot pressure in the right foot (Matsuda and Demura, 2013) indicating asymmetry also exists in static posture. A study by Hayafune et al. (1999) reported that body weight bearing of T has negative correlations with body weight bearing of metatarsals head. When foot touches the ground, the foot pronates and then supinates (Hutton and Dhanendran, 1979), which is indicated by the 1MT touches the ground earlier than both 3MT and 5MT. The supination distributes body weight pressure from 1MT laterally. This distribution is indicated by negative correlation of medio-lateral load as reported by Hayafune et al. (1999). While no significant differences was observed on 5MT, RH showed significantly greater CCF_{coeff} on 1MT at 3 km/h than LH, indicating more symmetrical pressure, followed by shorter time lag on 3MT at 1.5 and 3 km/h. From 3MT ground contact duration, it is obvious that the positive time lag in LH was caused by the right 3MT touched the ground earlier than the left 3MT. LH have asymmetrical timing of foot supination especially at lower speeds. The findings that walking speed affects contact duration of 1MT and 3MT to various degrees in both groups of handedness is in agreement with previous study on influence of walking speed on plantar pressures which found speed affected T, 1MT and 3MT, as well as heel, but not arc area and 5MT (Burnfield et al. 2004).

We did not find any significant difference on CCF_{coeff} of C both in LH and RH, indicating the pressure is relatively symmetrical. LH showed significantly longer LC ground contact duration than RC, indicating asymmetrical timing of ankle flexion. These data were also supported by time lags of CCF analysis, which showed that LH time lags were longer than RH in all speeds measured. All time lags were positive indicating that RC of LH reached the peak pressure earlier than the LC, while the LC has longer contact duration, which indicates the slower plantar flexion during transition from heel strike to push-off in left foot compared to right foot. Niu et al. (2011) who identified footedness only by ball kicking foot reported that dominant ankle had greater dorsiflexion angular velocities during two feet drop landing, which means in the dominant C had shorter contact duration. This is in disagreement with the result of our study which did not find its association with dominant side in both groups of handedness. Stance phase during gait cycle is divided into two phases, namely braking and propulsion phase (Neptune and Sasaki, 2005). LC longer contact duration in LH is associated with longer braking phase, representing control function, while the shorter RC contact duration represents propulsion function. However these findings are not related to dominant side, therefore in disagreement with propulsion-control hypothesis by Sadeghi et al. (2000).

TA muscle which has role in ankle dorsiflexion (Kendall et al., 1983) did not show any significant differences in all condition measured. On the other hand, Sol muscle which has role in ankle plantar flexion (Kendall et al., 1983) showed significant difference between LH and RH in time lag at 3 km/h. Soleus muscle of RH also showed significant increase of CCF_{coeff} between 1.5 and 4 km/h. Observation on interlimb coordination during walking indicates that ankle dorsiflexor half centres of homologous limbs inhibit each other, whereas the plantar flexor half centres are not coupled each other (van Hedel et al., 2002). The plantar flexor has been reported to be important during support, forward progression and swing initiation, and also contributes to anterior-posterior and vertical ground reaction force during propulsion phase (Neptune and Sasaki, 2005). Based on foot contact duration and lower limb muscles analyses, we can conclude that asymmetry occurs during propulsion phase, especially among LH.

ES showed increasing CCF coefficients from 1.5 km/h to 4 km/h in RH and significantly higher CCF coefficients compared to LH at 4 km/h. The significant increase of ES CCF_{coeff} resembles that of Sol muscle in RH. During walking, the peak activation of ES were found between contralateral heel strike and the onset of swing phase of ipsilateral leg, almost the same time with Sol muscle whose peak activation lies between push-off and onset of swing phase of ipsilateral leg (Ivanenko et al., 2004; Courtine et al., 2006). ES muscle main function during walking is to maintain postural stability by restricting excessive trunk movement especially in the frontal plane (Thorstensson, 1982). Bipedal humans walk with erect trunk on two legs, that is naturally unstable especially medio-laterally, therefore the development of lower limb movement and trunk vertical stability emphasizes the dynamic coordination of body balance and forward motion (Courtine et al., 2006). The humans foot is characterised with narrow width and longer length, and less stable compared to non-human primates such as chimpanzee which has wider width and abducted T (Kiriyama et al., 2005). However this foot shape with arc structure allows speedy progressions at the expense of medio-lateral stability (Kiriyama et al., 2005). As a study reported that limb dominance did not affect lower limbs kinematic and kinetic patterns during running (Brown et al., 2014), it seems that faster gait will increase gait symmetry. Special treatment on foot, especially between toes and metatarsals head, was also reported to have effect on ES muscle activation, since foot pronation affected internal rotation of the leg and ipsilateral pelvic tilt (Bird et al., 2003). We did not observe any significant changes in TB muscle activation in all conditions measured, indicating that the symmetrical activation of this muscle is less affected during walking with arm swing.

Laterality has been associated with asymmetries of cortical organization which is most distinctive in humans (Springer and Deutsch, 1997; Soros et al., 1999; Jung et al., 2003). The asymmetry also has been demonstrated in brain function such as language and attention (Steinmetz et al., 1991; Coren, 1992; Soros et al., 1999). Almost all RH were reported to have language lateralization on their left hemisphere, however the situation is less uniform in LH (Coren, 1992). Soros et al., (1999) observed that there is a strong correlation between handedness and the cortical hand representation among RH, but not in LH. While most RH are also right-footers, the situation is not clear for LH. We speculate that diverse results in gait asymmetry studies is related to limited number of left-handed subjects whose footedness have been largely neglected.

Animal locomotion is partially controlled by an intraspinal network of neurons capable of generating rhythmic movements of limbs called as central pattern generators (CPG) (Golubitsky et al., 1999). While CPG model has been assumed to be symmetrical (Golubitsky et al., 1999), during evolution, skilled hand movements evoked greater role of direct cortical-motoneuronal system which affects the degree of laterality in humans and non-human primates (Dietz, 2002). Greater symmetry due to increasing walking speed as observed especially in Sol and ES muscles of RH probably related to higher frequency of rhythmical movement thus greater role of CPG. However such effect was not observed in LH. Human locomotion is also controlled by central nervous system (CNS) by utilising a set of activation patterns distributed to several different muscles which produced output during phases of motor task based on both feed forward and feedback signals from the dynamic condition of the whole limbs (Lacquaniti et al., 2012; Oliveira et al., 2013). We speculate that there is difference in degree of direct cortical-motoneuronal and intraspinal intervention during locomotion between LH and RH.

Main limitation in the studies of laterality is limited number of non-right-handed subjects, thus most studies tried to generalised from right-handed and right-footed subjects. This study shows that such a generalisation is not applicable to left-handed and mixed-footed subjects whose asymmetrical features are different from right-handed subjects, but not associated with sidedness. Laterality is not always constant from the birth and about 10% of each LH and RH was found switched into the opposite handedness (Coren, 1992). During walking, bilateral muscle activation patterns were observed (MacLellan et al., 2014) and asymmetrical interlimb learned ankle movement was reported (Morris et al., 2009). We compared the difference between two different groups of laterality but did not try to generalize the correlation between degree of laterality and walking symmetry. In order to draw a general conclusion on laterality during human evolution, the availability of various subjects is important, and future study should avoid generalising any finding in right-lateralised subjects in other laterality groups.

5.5 Conclusions

During walking, RH and LH showed different asymmetry level which is not in mirrored characteristics. LH showed asymmetry in calcaneus ground contact duration, whereas RH did not show any asymmetry in all footsole points measured. LH also showed greater asymmetry in pressure of T and 3MT pressure timing at 1.5 and 3 km/h. RH showed that increasing walking speed from 1.5 km/h to 4 km/h improve symmetry in Sol and ES muscle activation. These results will be useful for various studies, such as therapy for improving gait symmetry, or in general view to understand the difference between RH and LH where previous studies reported mixed results.

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Chapter 6
The comparison between manual pushing and walking without hand force exertion based on symmetry effect as measured by foot pressure and muscle activation

6.1 Introduction

Symmetry is a phenomenon which is very common in the universe. Symmetry during gait has been very important in rehabilitation therapy of stroke patients (Dickstein et al., 1986; Norman et al., 1995; Richards and Olney, 1996; Laufer et al., 2001; Saiburg and Duff, 2006). Gait features such as velocity, cadence, step length, stride length, and stride length to lower extremity ratio are excellent tools for assessing physical therapy assessment among hemiparetic subjects (Holden et al., 1984; von Schroeder et al., 1995). Gait training on treadmill has been applied for stroke survivor rehabilitation (Waagfjord et al., 1990; Norman et al., 1995; Laufer et al., 2001). One of the most common walking conditions during therapy is the patient walking with both hands holding a handlebar on the treadmill to help them to avoid fall during walking.

On the other hand, in normal people, one representative of symmetry measure is laterality which refers to asymmetrical preferential use of limbs and sensory (Schneiders, et al. 2010). In evolutionary studies, bipedalism has been associated with the development of laterality. Study on chimpanzees found that a bipedal stance without hand support will evoke hand preferences (Braccini, et al. 2010). Study on Sichuan snub-nosed monkeys also found that foot preference was significantly stronger in bipedal action (Zhao, et al. 2008). About 85% of human population are right-handed (Uomini, 2006). Other suggested around 90% of humans are right handed, around 80% are right footed and around 70% are right eyed (Carey, et al. 2001). Based on population studies, right foot tends to be the preferred foot of right-handers, however the situation remains unclear for left-handers (Chibber and Singh, 1970; Chapman, et al. 1987; Peters, 1988).

Laterality can also be changed due to pathological cause or cultural. Jones (1870) introduced the word stammerer to call people whose laterality has been transferred to the opposite side. The study by Jones (1870) reported that 4 % of the population are transferred, of which 1% of born right-handers were transferred to left-handers by accident, 1% of born left-handers were transferred to right-handers by accident, and 76% of born left-handers were transferred to right-handers by purposive means such as cultural education. More recent study by Coren (1992) reported that natural handedness in the population is 90% right-handers and 10% left-handers, of which 10% of the population of both handedness were transferred to the opposite handedness due to pathological reason.

Left-footers were found to control their unipedal posture differently from right-footers (Golomer and Mbongo, 2004). Hart and Gabbard (1997) suggested that lower limb choice for postural stabilization in bipedal context may be independent of that in unipedal, where dominant leg is used for more demanding task. Other study reported that there was no lateral dominance during static single-leg stance, however higher frequency of movement was found to strengthen lateral dominance for postural stability (Kiyota and Fujiwara, 2014). A study on footedness during walking using footprint method found that footedness does not affect gait parameters (Zverev, 2006). In study of gait initiation, Hesse et al. (1997) found that in normal subjects gait initiation is highly symmetrical. This suggestion was in disagreement with research review from Sadeghi, et al. (2000) who hypothesized asymmetrical lower limbs behaviour during gait as a reflection of natural functional differences in propulsion and control. This was supported by Dessery et al (2011) who observed asymmetrical frontal body motion influenced by footedness in gait initiation. Herzog, et al. (1989) measured normal human gait using symmetry index found that gait asymmetries were larger than expected.

In the study of symmetry, there is a paradoxical phenomenon called as symmetry breaking where the symmetric system starts to behave less symmetrically, when the symmetry of the resulting state of the system is a subgroup of the symmetry group of the whole system (Stewart and Golubitsky, 2011). During walking, both legs move half a period out of phase which shows symmetry breaking. In gait, two types of symmetry are observable, namely spatial symmetry and spatio-temporal symmetry, in bipeds, only the latter is observable (Golubitsky et al., 1999).

This study investigated the difference between three gait conditions: manual pushing while walking, walking with arm-swing, and walking with both hands hold on to a handlebar to help gait stability without any hand force exertion, on gait spatio-temporal symmetry by measuring muscular activation and foot pressure. The results of this study will provide information in difference between normal walking with arm-swing with pushing while walking and walking holding condition which is commonly used in gait therapy.

6.2 Methods

6.2.1 Subjects

Subjects were 17 healthy young adult males obtained from Chiba University student population (age 28 $\pm$ 5 years; height 169.9 $\pm$ 6.9 cm; weight 64.6 $\pm$ 7.3 kg). The laterality data of subjects were measured by Waterloo Footedness Questionnaire (WFQ) and Waterloo Handedness Questionnaire (WHQ) (Elias, et al. 1998), where 11 subjects were found to be both right-handed (50.25 $\pm$ 12.42) and right-footed (9.25 $\pm$ 4.13), and 6 subjects were grouped into left-handers (-15.17 $\pm$ 13.7) and mixed-footers (0.0 $\pm$ 7.5). To ease grouping we call the first group as right-handers (RH) and the second groups as left-handers (LH). Prior to the experiment, subjects gave informed consent which was approved by the Ethics Committee of the Graduate School of Engineering, Chiba University (25-25).

6.2.2 Instruments

Electromyogram (EMG) data were collected bilaterally from four muscles: tibialis anterior (TA), soleus (Sol), lumbar erector spinae (ES) and triceps brachii (TB). The strain amplifier output and EMG electrodes were connected to a Biopac MP 150 data acquisition system (Biopac Systems, USA), which then was connected to a personal computer. Ten FSR-400 pressure sensors (Interlink Electronics, USA) were attached bilaterally on great toe, 1st metatarsal, 3rd metatarsal, 5th metatarsal, and calcaneus adapted from a method used by Kiriyama et al. (2005) in order to measure foot pressure and contact duration. All subjects wore the same footwear to standardise friction. Each trial period was determined by a visual display timer and light sensor (Kodenshi Corp., Japan).

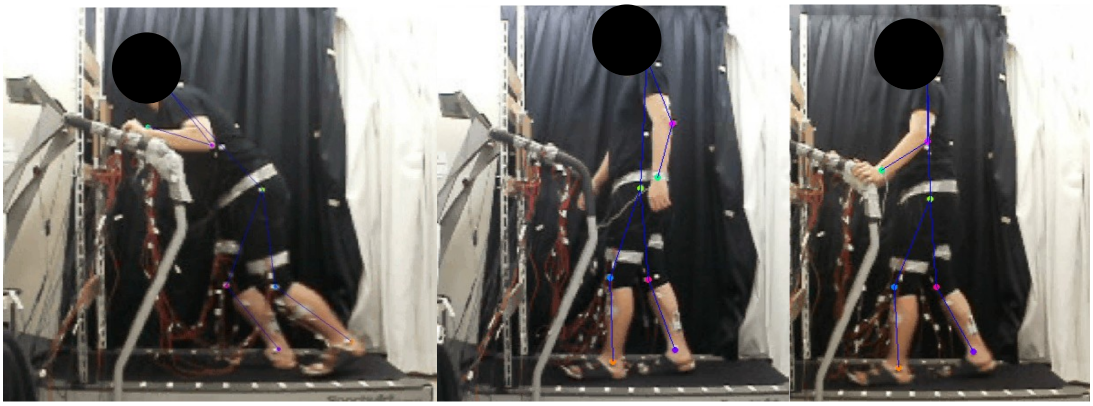

Figure 6-1. Three conditions of gait during experiment: pushing while walking, walking with arm-swing and walking with hands holding a handlebar without any force exertion on a treadmill.

6.2.3 Experiment Procedure

Subjects performed the three conditions of gait: pushing while walking, walking with arm-swing and walking with hands holding a handlebar without any force exertion on a treadmill (SportsArt Fitness, Taiwan) in three speeds: 1.5 km/h, 3 km/h, and 4 km/h. For manual pushing condition, subjects were required to exert at least 50% of their maximum static pushing force. Before the main trials, subjects were familiarised with treadmill walking. Each walking and walking holding condition trial was performed in 1 minute of which 10 seconds most consistent middle part was recorded. Manual pushing trial was performed in 10 seconds during subjects walking for around 1 minute. Each trial condition was performed three times. All conditions of gait and all speeds order were randomized. A three minutes rest was given between trials. During trial, subjects were instructed to gaze at the monitor which display timer. Gazing on a monitor is advantageous, because it prevents asymmetrical visual influence on balance control as reported in previous studies (Bessou, et al., 1999; Golomer and Mbongo, 2004; Nagano, et al., 2006).

6.2.4 Data Analysis

Data analysis was based on one gait cycle by using foot pressure sensor as reference. We divided data into right and left side gait cycle, where data from left side muscles and pressure sensor were analysed according to left foot gait cycle, and data from the right side were analysed according to right foot gait cycle. EMG and pressure sensor data were collected at a 1000-Hz sampling rate. Raw EMG signals were band-pass filtered between 15-250Hz, and root mean square (RMS) was derived. All data were normalized into 100 data points in one gait cycle. From the 10 seconds recorded data, 3 gait cycles were taken into analysis.

To measure spatio-temporal change in symmetry of variables measured in left and right side, cross-correlation function (CCF) was employed. The coefficients vary between -1 and +1, where a positive correlation value indicates the signals are in phase, and a negative value indicates inverse relationship (Nelson-Wong et al. 2009). CCF also measured time lag (τ) between the two signals. Data measured from the left and right sides were arranged so that if the peak of right side were earlier than the left side, time lag will be positive, while if it were later, time lag will be negative.

In statistical analyses, subjects were grouped into right-handers (RH) and left-handers (LH) groups. All recorded data as well as cross-correlation coefficient data were averaged between subjects in the same group. We analysed the characteristics of each group, and then for Cross-correlation coefficient and time lag data we also performed comparison between groups. The Shapiro-Wilk test was used to determine whether parametric or non-parametric method should be used. We employed Student`s paired T-test for parametric method and Wilcoxon signed-rank test for non-parametric method in order to compare data between left and right side. Data comparison between different velocities in the same group was performed using one-way repeated measures ANOVA with Bonferroni post-hoc test for parametric method and Friedman Test with the Wilcoxon signed-rank post-hoc test for non-parametric method. Because the two groups of subjects have unequal samples, in order to compare both groups, independent T-test for parametric method and Mann-Whitney U test for non-parametric method were used. Statistical significance was set at $p<0.05$. All statistical analyses were performed with Microsoft Excel 2010J and IBM SPSS 17J.

6.3 Results

6.3.1 Gait cycle duration

In gait cycle duration, LH did not show any significant differences between gait condition in all speeds measured ($p<0.05$). On the other hand, RH showed that gait condition affected gait cycle duration, as shown in Figure 6-2.

At 1.5 km/h walking speed, Friedman test revealed significant differences on left foot gait cycle duration between pushing, walking and walking holding condition ($X^2(2)=9.600$, $p=0.008$). Post-hoc analysis with Wilcoxon signed-rank tests with a Bonferoni correction applied found that walking holding condition has significantly longer gait cycle duration on left foot of RH than both pushing ($Z=-2.803$, $p=0.005$) and walking ($Z=-2.090$, $p=0.037$). The speed of 1.5 km/h also revealed such condition on right foot gait cycle, where walking holding condition also has longer gait cycle duration than both pushing ($p=0.047$) and walking ($p=0.009$).

At 3 km/h walking speed, such phenomena were also observed. Friedman test conducted found significant difference in gait cycle duration on left foot between pushing, walking and walking holding condition ($X^2(2)=7.800$, $p=0.02$). Post-hoc analysis with Wilcoxon signed-rank tests found that walking holding condition has significantly longer gait cycle duration on left foot of RH than both pushing ($Z=-2.497$, $p=0.013$) and walking ($Z=-2.599$, $p=0.007$). Friedman test also revealed significant difference in gait cycle duration on right foot between pushing, walking and walking holding condition ($X^2(2)=7.800$, $p=0.02$), with Wilcoxon signed-rank test found that walking holding condition has significantly longer gait cycle duration on left foot of RH than both pushing ($Z=-2.293$, $p=0.022$) and walking ($Z=-2.497$, $p=0.013$).

At 4 km/h walking speed, one way ANOVA with repeated measures found significant difference only on right foot, where walking holding condition have significantly longer gait cycle duration that pushing ($p<0.05$).

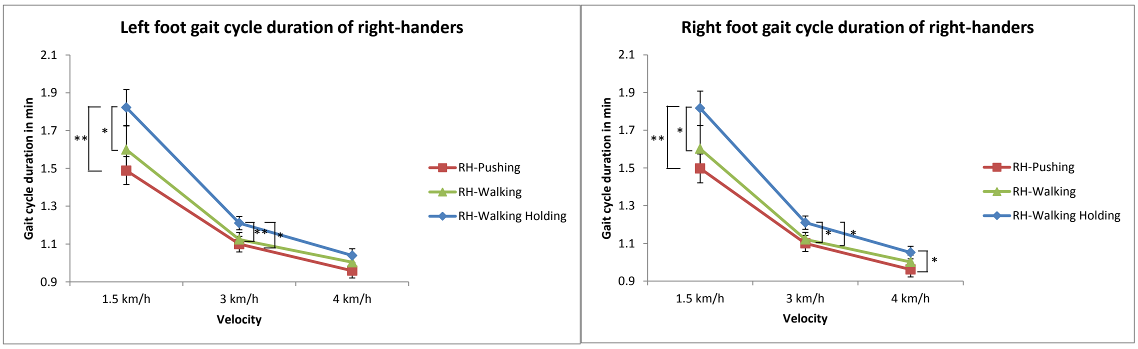

Figure 6-2. Left foot (left) and right foot (right) gait cycle duration of RH in pushing, walking and walking holding condition (**p<0.01; *p<0.05).

6.3.2 Foot contact duration

As shown in Figure 6-3, Stance phase duration relative to gait cycle of LH seems to be shorter with increasing walking speed in walking and walking holding condition in both left and right foot, but not in pushing condition which seems to increase at 4 km/h. However one way ANOVA with repeated measure did not find any significant difference between gait conditions in all speeds measured in both feet (p>0.05).

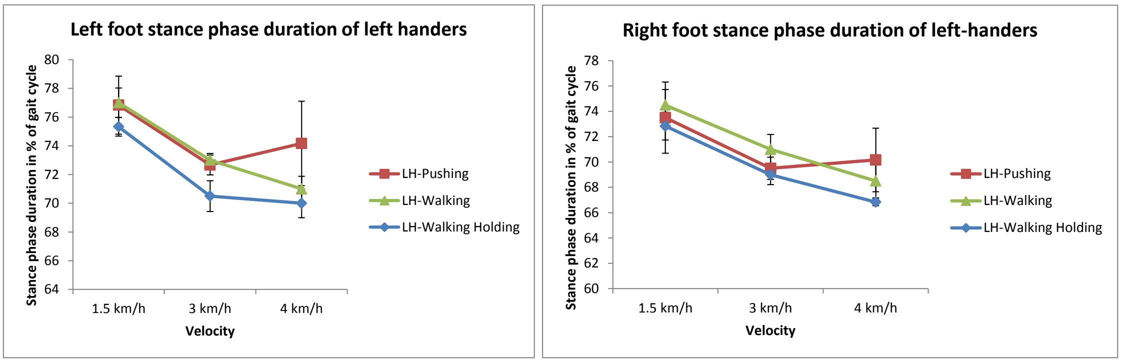

Figure 6-3. Left foot (left) and right foot (right) stance phase duration of LH.

As in LH, in RH, from general data observation, stance phase duration relative to gait cycle of RH seems to be shorter with increasing walking speed in walking and walking holding condition in both left and right foot, but not in pushing condition which seems to increase at 4 km/h on the left foot of RH (Figure 6-4). One way repeated measures ANOVA with post hoc test using Bonferroni correction revealed that at 4 km/h, pushing resulted in significantly longer stance phase duration on left foot of RH than both walking (p=0.032) and walking holding (p=0.022).

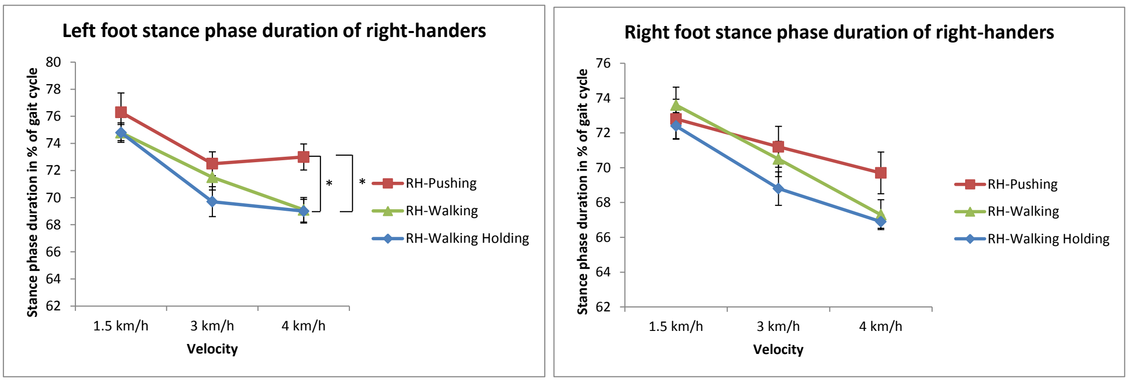

Figure 6-4. Left foot (left) and right foot (right) stance phase duration of RH (*p<0.05).

In accordance with symmetry breaking concept, we also measured ground contact duration of five points of footsole: great toe, 1st metatarsal, 3rd metatarsal, 5th metatarsal, and calcaneus. This measurement revealed the effect of different gait condition to various parts of footsole which contribute to ankle biomechanics in general.

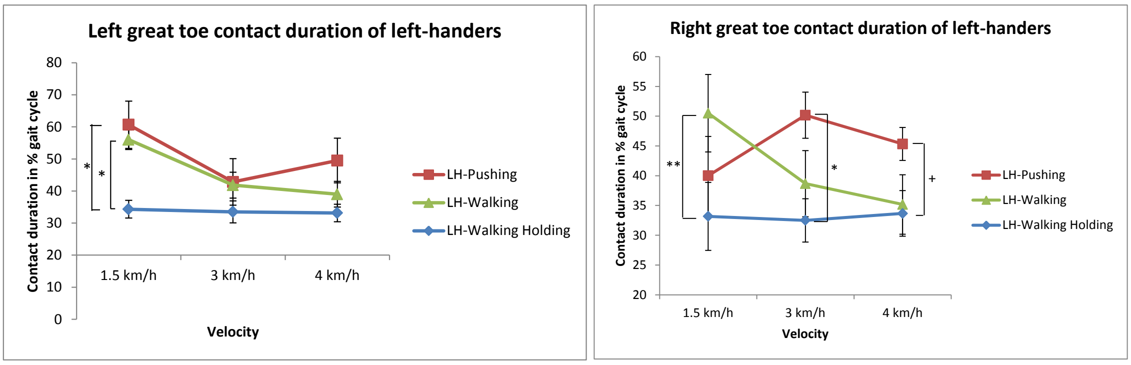

Figure 6-5. Ground contact duration in one gait cycle of LH: left great toe (left) and right great toe (right) (**p<0.01; *p<0.05; ^{+}p<0.1).

Figure 6-5 shows that Friedman test followed with Wilcoxon signed-rank test found left great toe of LH generated significantly shorter contact duration at 1.5 km/h walking speed in walking holding than both pushing (Z=-1.992, p=0.046) and walking (Z=-2.201, p=0.028). Right great toe of LH showed significantly shorter ground contact duration in walking holding condition than walking at 1.5 km/h (p=0.003) and pushing at 3 km/h (p=0.03). In general, both left and right great toe contact duration was relatively unchanged by increasing speed in walking holding condition compared to pushing and walking.

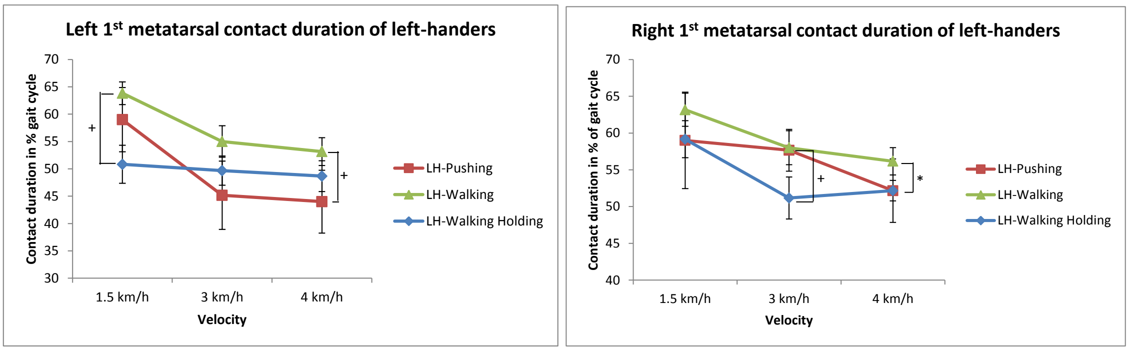

Figure 6-6. Ground contact duration in one gait cycle of LH: left 1st metatarsal (left) and right 1st metatarsal (right) (**p*<0.05; $^{+}$*p*<0.1).

LH did not show any significant difference due to gait condition in left 1st metatarsal ground contact duration (*p*>0.05). Figure 6-6 (right) shows that at 4 km/h walking speed, right 1st metatarsal of LH has longer ground contact duration during walking than walking holding (*p*=0.04). We did not observe any significant difference between pushing, walking and walking condition in bilateral ground contact duration of 3rd metatarsal, 5th metatarsal, and calcaneus of LH (*p*>0.05).

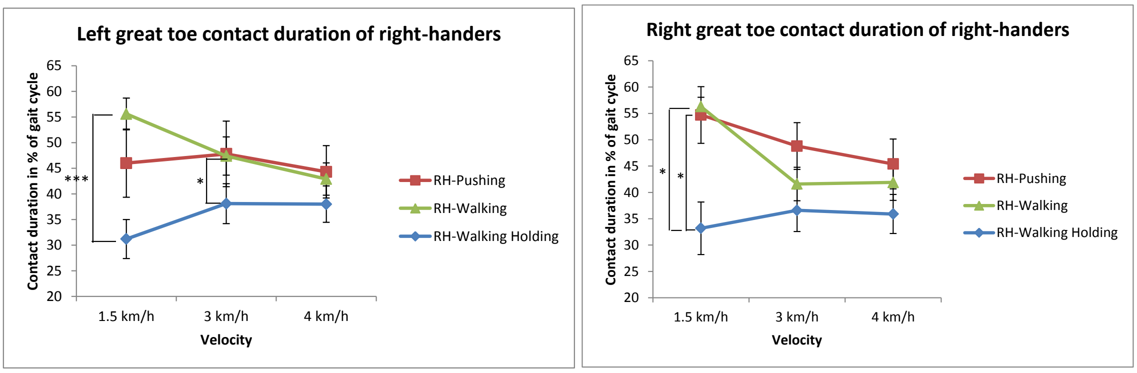

Figure 6-7. Ground contact duration in one gait cycle of RH: left great toe (left) and right great toe (right) (****p*<0.001; **p*<0.05; $^{+}$*p*<0.1).

Figure 6-7 shows great toe ground contact duration of RH. As shown in Figure 6-7 (left), at both 1.5 and 3 km/h, left great toe of RH has significantly shorter ground contact duration during walking holding than walking (*p*<0.001 and *p*<0.05, respectively). One way repeated measure ANOVA with post hoc test using Bonferroni correction revealed that walking holding resulted in significantly shorter ground contact duration than pushing (p=0.011) and walking (p=0.013) on right great toe of RH at 1.5 km/h walking speed (Figure 6-7-right).

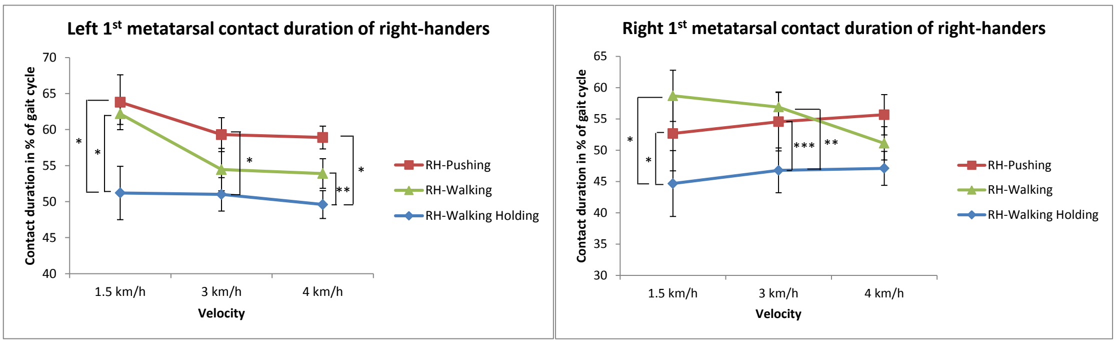

Figure 6-8. Ground contact duration in one gait cycle of RH: left 1st metatarsal (left) and right 1st metatarsal (right) (***$p<0.001$; **$p<0.01$; *$p<0.05$).

Left 1st metatarsal of RH showed that walking holding condition resulted in significantly shorter ground contact duration than both pushing (p=0.038) and walking (p=0.017) at 1.5 km/h, than pushing at 3 km/h (Z=-2.314, p=0.021), as well as than both pushing (Z=-2.492, p=0.013) and walking (Z=-2.673, p=0.008) at 4 km/h (Figure 6-8-left). Friedman test found significant difference between ground contact duration of right 1st metatarsal of RH in the three gait conditions at 1.5 km/h ($X^2(2)$=7.588, p=0.023). Post-hoc analysis with Wilcoxon signed-rank tests with a Bonferoni correction applied found that at 1.5 km/h, walking holding resulted in significantly shorter contact duration in the right 1st metatarsal of RH than pushing (Z=-2.033, p=0.042) and walking (Z=-2.549, p=0.011). One way repeated measure ANOVA with post hoc test using Bonferroni correction revealed that at 3 km/h, walking holding also resulted in significantly shorter ground contact duration of right 1st metatarsal of RH than pushing ($p<0.001$) and walking ($p<0.01$).

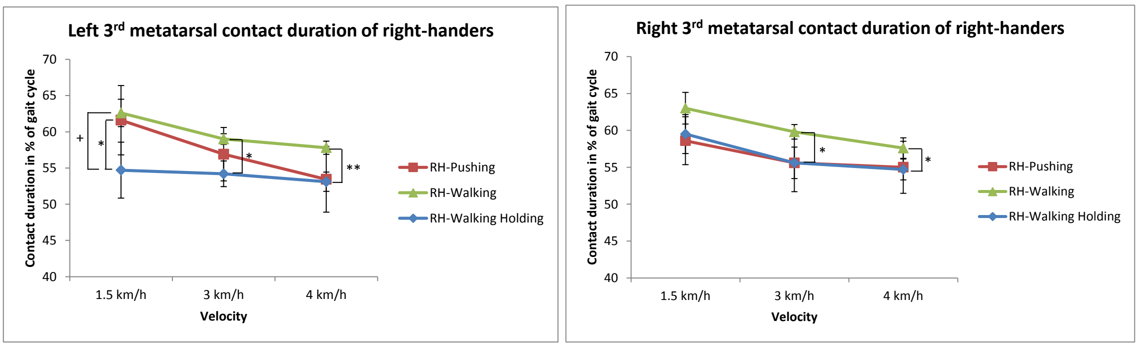

Figure 6-9. Ground contact duration in one gait cycle of RH: left 3rd metatarsal (left) and right 3rd metatarsal (right) (**p<0.01; *p<0.05; ^{+}p<0.1).

As shown in Figure 6-9 (left), one way repeated measure ANOVA with post hoc test using Bonferroni correction found that left 3rd metatarsal of RH generated significantly shorter ground contact duration during walking holding condition than pushing at 1.5 km/h (p=0.04), and walking at both 3 km/h (p=0.013) and 4 km/h (p<0.01). Figure 6-9 (right) shows that right 3rd metatarsal of RH generated shorter ground contact duration during walking holding than walking at both 3 km/h (p<0.05) and 4 km/h (Z=-2.530, p=0.011). We did not observe any significant difference on ground contact duration of RH's bilateral 5th metatarsal (p>0.05).

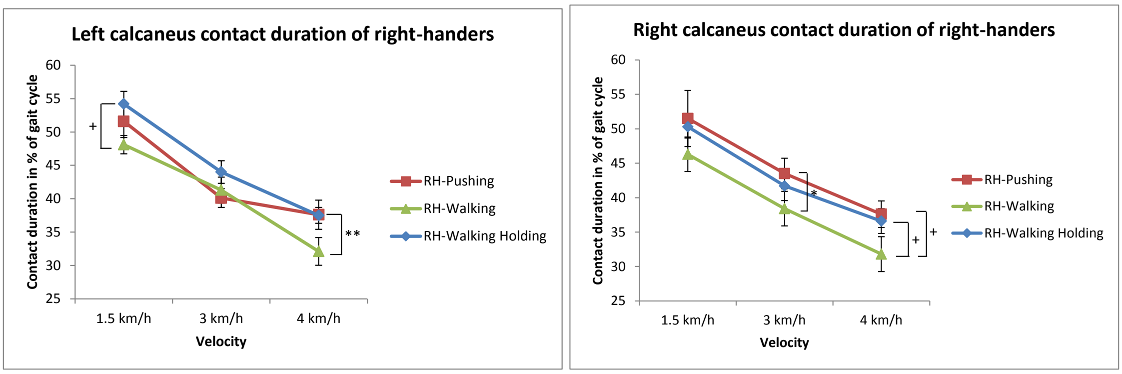

Figure 6-10. Ground contact duration in one gait cycle of RH: left calcaneus (left) and right calcaneus (right) (**p<0.01; *p<0.05; ^{+}p<0.1).

As shown in Figure 6-10, RH generated significantly shorter ground contact duration on their left calcaneus during walking at 4 km/h than walking holding at the same speed (p<0.01), and on their right calcaneus during walking at 3 km/h than pushing at the same speed (p<0.05). Left calcaneus only showed tendency of shorter ground contact duration during walking compared to walking holding ($p<0.1$), whereas right calcaneus only showed tendency of shorter ground contact duration during walking compared to both pushing and walking holding at 4 km/h ($p<0.1$).

6.3.3 Cross-correlation function analyses

In observation on cross-correlation coefficient of five points of footsole, we did not find any significant difference between gait conditions at all speeds measured in 3rd metatarsal and calcaneus ($p>0.05$). As shown in Figure 6-11, LH showed that pushing resulted in significantly less cross-correlation coefficient on great toe compared to both walking ($p<0.01$) and walking holding condition ($p<0.05$) at 3 km/h. RH, on the other hand, showed that walking holding resulted in greater symmetry compared to both pushing ($p<0.05$) and walking ($p<0.05$). We did not observe any significant difference in time lag of great toe in both LH and RH ($p>0.05$). Walking condition resulted significant difference between LH and RH in cross-correlation coefficient at 1.5 and 3 km/h, and in time lag at 3km/h ($p<0.05$). However, in two other gait conditions, there was no significant difference observed between LH and RH.

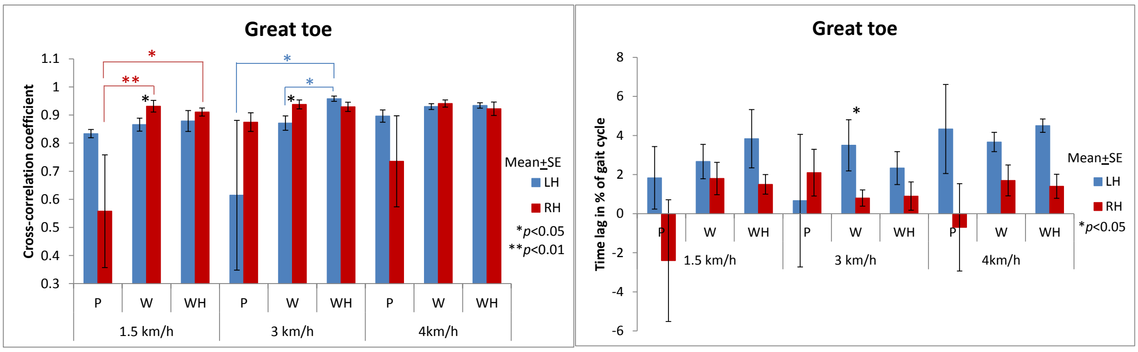

Figure 6-11. Cross-correlation coefficient (left) and time lag (right) of great toe (**$p<0.01$; *$p<0.05$).

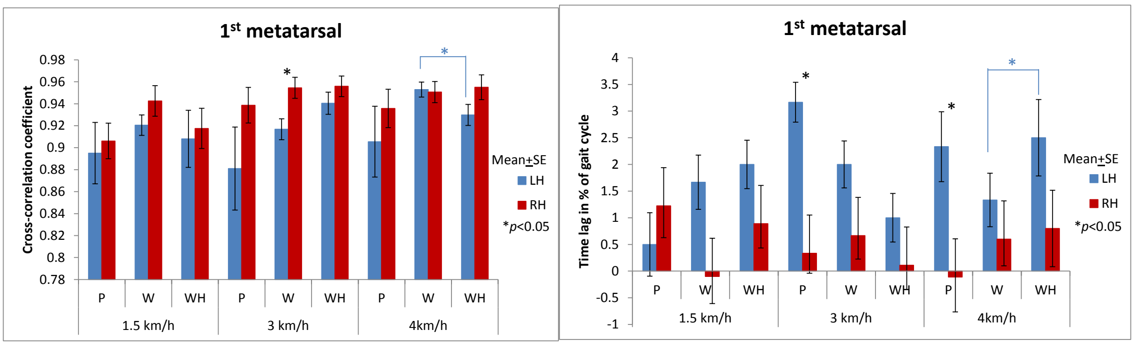

Figure 6-12. Cross-correlation coefficient (left) and time lag (right) of 1st metatarsal (**p*<0.05).

As shown in Figure 6-12 (left), from all conditions measured, only LH demonstrated that at 4 km/h speed, walking resulted in greater cross-correlation coefficient of 1st metatarsal ground pressure than walking holding condition ($p<0.05$). This data is accompanied by significant difference in time lag between walking which has shorter time lag than walking holding ($p<0.05$) as shown in Figure 6-12 (right). RH did not show any significant difference in all conditions measured ($p>0.05$). We did not observe any significant difference in time lag of 1st metatarsal in both LH and RH ($p>0.05$). Walking resulted significant difference in cross-correlation coefficient between LH and RH at 3 km/h ($p<0.05$) where RH showed more symmetrical 1st metatarsal pressure, while pushing generated significant difference between LH and RH in time lag at 3 and 4 km/h, where LH showed longer time lag than RH ($p<0.05$).

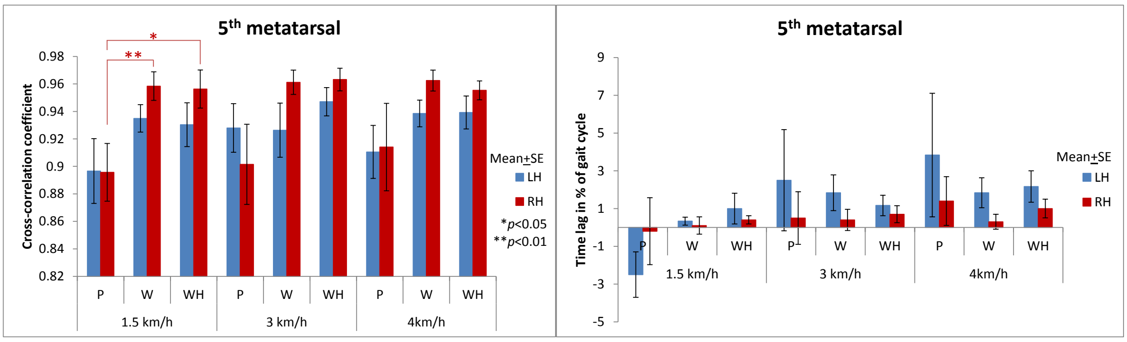

Figure 6-13. Cross-correlation coefficient (left) and time lag (right) of 5th metatarsal (LH: mean-SE; RH: mean+SE) (***p*<0.01; **p*<0.05).

As shown in Figure 6-13, significant difference between gait conditions was only observed in RH at 1.5 km/h where manual pushing resulted in lower cross-correlation coefficient than walking (p<0.01) and walking holding (p<0.05). We did not observe any significant difference in time lag of 5th metatarsal in both LH and RH (p>0.05).

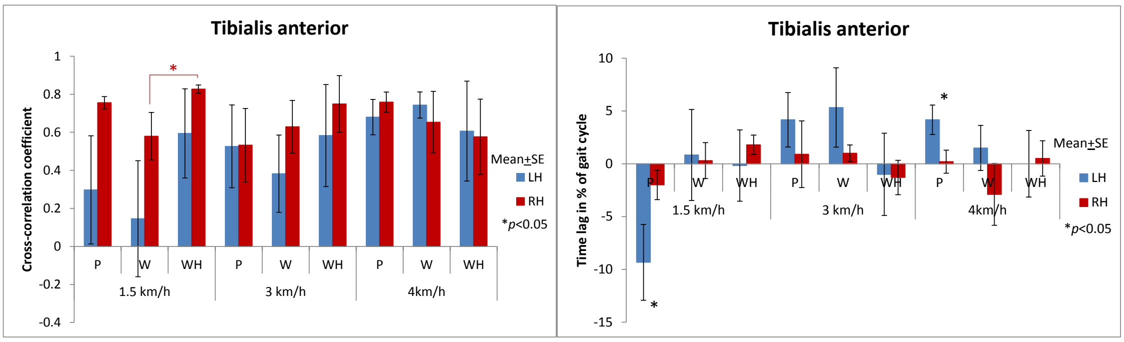

Figure 6-14. Cross-correlation coefficient (left) and time lag (right) of tibialis anterior muscle (LH: mean-SE; RH: mean+SE) (*p<0.05).

Figure 6-14 shows that significant difference in cross-correlation of tibialis anterior muscle activation only occurred between walking and walking holding in RH at 1.5 km/h, where walking holding resulted in greater cross-correlation coefficient (p<0.05). We did not observe any significant difference in time lag of tibialis anterior muscle activation in both LH and RH (p>0.05). In addition, as has been described in Chapter 4, manual pushing at 4 km/h resulted in significantly longer time lag in LH compared to RH (p<0.05).

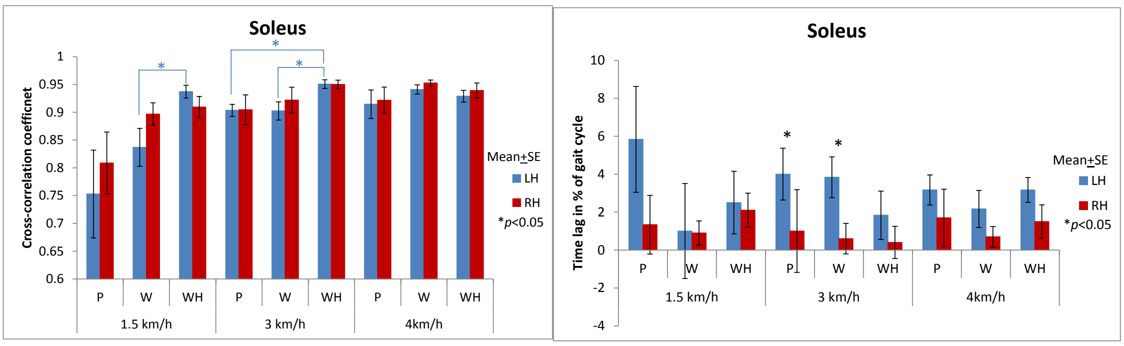

Figure 6-15. Cross-correlation coefficient (left) and time lag (right) of soleus muscle (LH: mean-SE; RH: mean+SE) (*p<0.05).

Significant difference in soleus muscle activation was only observed in LH. Walking holding showed significantly greater cross-correlation coefficient in soleus muscle activation than walking (p<0.05) at 1.5 km/h. Among LH, walking holding also resulted in significantly greater cross-correlation coefficient than both manual pushing (p<0.05) and walking (p<0.05) at 3 km/h (Figure 6-15). We did not observe any significant difference in time lag of soleus muscle activation in both LH and RH (p>0.05). At 3 km/h, both pushing and walking resulted in significantly longer time lag of soleus muscle activation in LH than RH (p<0.05), as have been described in Chapter 4 and 5.

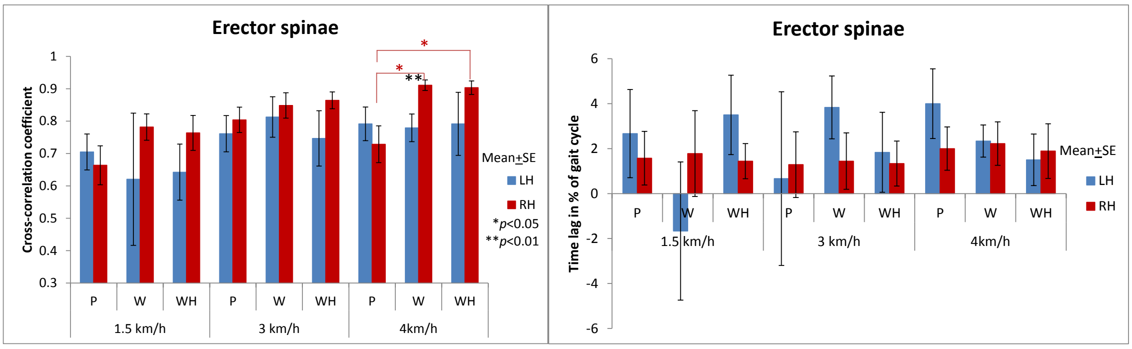

Figure 6-16. Cross-correlation coefficient (left) and time lag (right) of erector spinae muscle (LH: mean-SE; RH: mean+SE) (*p<0.05).

Figure 6-16 shows that significant difference in cross-correlation coefficient of erector spinae muscle activation due to different gait condition was only observed in RH at 4 km/h, where based on cross-correlation coefficient, manual pushing resulted in significantly lower symmetry than walking with arm-swing (p<0.05) and walking holding (p<0.05). We did not observe any significant difference in time lag of erector spinae muscle activation in both LH and RH (p>0.05). Interestingly, as described in Chapter 5, walking at higher speed of 4 km/h generated more asymmetrical erector spinae muscle activation in LH compared to RH, as demonstrated by significant difference of cross-correlation coefficient (p<0.05). This finding was not observed in other conditions.

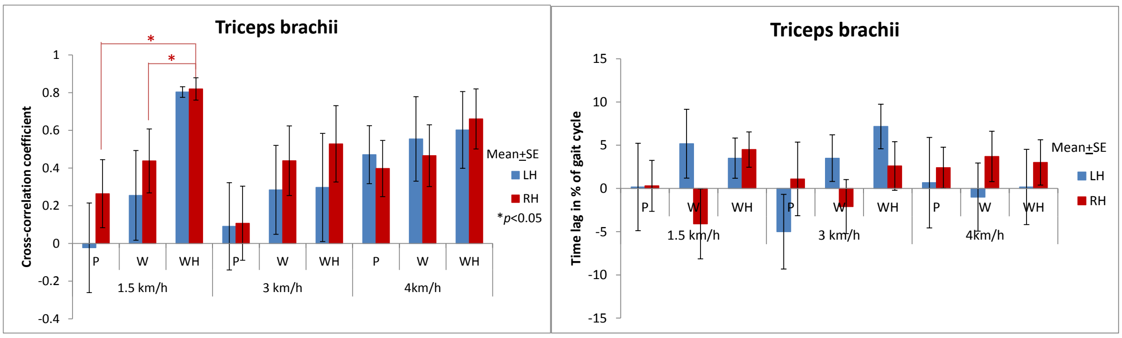

Figure 6-17. Cross-correlation coefficient (left) and time lag (right) of triceps brachii muscle (LH: mean-SE; RH: mean+SE) (*:p<0.05).

As shown in Figure 6-17, significant difference in cross-correlation coefficient of triceps brachii muscle activation was only observed in RH at 1.5 km/h, where manual pushing showed lower cross-correlation coefficient than walking with arm swing (p<0.05) and walking holding condition (p<0.05), representing lower symmetry. We did not observe significant difference in time lag of triceps brachii muscle activation (p>0.05).

6.4 Discussion

LH did not show any significant difference in gait cycle duration in all conditions measured. On the other hand, RH showed that manual pushing shortened gait cycle duration, especially at lower speed at 1.5 and 3 km/h. When walking with carrying load, stride length was shortened and stride frequency was increased in order to reduce the support phase of the metatarsal layer system that is used in the push-off phase of backpack walking (Kinoshita, 1985; Pascoe et al., 1997). Nottrodt and Manley (1989) supported the suggestion but such condition should be above preferred maximum load. In our study, this condition was demonstrated with shorter gait cycle duration, which means higher stride frequency. Our study revealed that this strategy was adopted by RH who represent the majority of population, but not adopted by LH to cope with greater load.

In stance phase duration, LH did not show any significant difference between pushing, walking and walking holding. However, at faster gait such at 4 km/h left foot of RH showed that stance phase duration during pushing is significantly longer than during walking and walking holding condition. Pang and Yang (2000) suggested that stance phase is prolonged and swing phase is delayed when the load on the limb is increased. LH did not show that they significantly adopted this strategy in the measured speeds, unlike RH. This finding also suggested that report by Lacquaniti et al. (2012) that stance phase duration has negative correlation with speed during walking is not applied for pushing.

Unlike suggestion by Kinoshita (1985) that ground contact duration of metatarsals head should be reduced with increasing load, this study showed various results depending on gait condition and laterality group. LH only showed longer ground contact duration on right 1st metatarsal during walking than walking holding, while no significant difference was observed on left 1st metatarsal, and bilateral 3rd and 5th metatarsal. Compared to pushing and walking, walking holding condition resulted in significantly shorter ground contact duration on both left and right 1st and 3rd metatarsal of RH. In general, walking holding also resulted in lower ground contact duration on bilateral great toe of both LH and RH. Considering that walking holding yield less load on the foot than pushing, our study is in disagreement with the study by Kinoshita (1985). If we assume that walking holding yield less load than both pushing and walking, ground collision-energy loss compensation strategy proposed by Kuo (2002) and Bertram and Hasaneini (2013) probably provide an explanation. Walking is one of the most energetically demanding daily activities, and performing push-off pre-emptively at the foot of the trailing limb before the heel-strike of the next stance foot can reduce energy loss up to 66%. This could be achieved by lengthening posterior foot contact duration of the trailing limb. Walking holding as a less energetically demanding gait does not require such compensation considering ground collision loss is probably also compensated by the hand that holding the handlebar.

Unlike LH who did not show any significance, RH showed significantly shorter ground contact duration on their left calcaneus during walking at 4 km/h than walking holding at the same speed, and on their right calcaneus during walking at 3 km/h than pushing at the same speed. The sole of the foot under the heel is covered by a layer of subcutaneous connective tissue up to 2 cm thick with a system of pressure chambers that both acts as a shock absorber and stabilises the sole in order to distribute the compressive forces exerted on the heel during gait (Schulte and Schumacher, 2006). In walking, calcaneus stresses did not reach its peak during heel-strike, but were generated late in the stance phase (Giddings et al., 2000). Gait condition probably affected the timing of peak calcaneus stresses on RH but not on LH.

LH showed that walking holding resulted in greater symmetry in great toe than manual pushing and walking. At low speed such as 1.5 km/h, RH showed that manual pushing generated less symmetrical pressure on great toe and 5th metatarsal compared to walking with arm-swing and walking holding condition. This indicates that force exertion of both arms creates asymmetry on those two parts of the foot. The main function of great toe is to support the metatarsal heads to take the full body weight when the heel is raised (Lambrinudi, 1932; Hughes et al., 1990). Body weight bearing of great toe has negative correlations with body weight bearing of metatarsals head (Hayafune et al., 1999). After heel strike, metatarsals head touches the ground earlier than great toe. In metatarsals head itself, when the foot touches the ground the foot pronates than supinates (Hutton and Dhanendran, 1979), which is shown by 1st metatarsal touches the ground earlier than 5th metatarsal. These data indicate that RH undergoes asymmetrical foot pressure during later stage of stance phase, especially during pre-swing. On the contrary, LH showed greater symmetry in 1st metatarsal pressure at 4 km/h during walking with arm-swing compared to walking holding, supported by lower time lag during walking with arm-swing compared to walking holding at 4 km/h. These data demonstrate that LH, at higher speed showed higher symmetry during pronation following plantar flexion after heel-strike in walking with arm-swing.

At low speed such as 1.5 km/h, RH showed that walking holding condition improves tibialis anterior muscle activation symmetry compared to walking with arm-swing. Tibialis anterior has an important role in dorsiflexion of ankle (Kendall et al., 1983) suggesting that at low speed among RH, walking holding is useful to improve symmetry in ankle dorsiflexion. Considering time lag of tibialis anterior muscle activation did not change due to gait condition, walking holding condition improves symmetry magnitude of activation rather than activation timing.

In soleus muscle activation, unlike tibialis anterior which were more affected in RH, gait condition only affected LH soleus muscle activation symmetry. Walking holding condition generated more symmetrical soleus muscle activation compared to walking with arm-swing at both 1.5 and 3 km/h, and compared to manual pushing at 3 km/h.

Soleus muscle has role in plantar flexion of the ankle (Kendall et al., 1983). Plantar flexor muscles are important during support, forward progression, and swing initiation (Neptune and Sasaki, 2005). Walking holding improves symmetry of soleus muscle activation among LH during swing initation or propulsion phase at 1.5 and 3 km/h.

Erector spinae muscle main function during walking is to maintain postural stability (Thorstensson, 1982). Bipedal humans with erect trunk on two legs that is naturally unstable (Courtine et al., 2006), with foot whose characteristics is narrow width and longer length which less stable medio-laterally (Kiriyama et al., 2005). RH showed that manual pushing generated less symmetrical erector spinae muscle activation at 4 km/h compared to walking with arm-swing and walking holding. This finding indicates that trunk was less stable during manual pushing than the two other conditions, especially due to requirement to transfer force from ground to the pushed object.

Triceps brachii muscle has role in extending elbow joint and additionally assist in adduction and extension of the shoulder joint (Kendall et al., 1983). This study found that in RH, pushing at low speed such as 1.5 km/h generated triceps brachii muscle activation less symmetrically compared to walking with arm-swing and walking holding condition. The asymmetry of triceps brachii muscle activation during pushing at low speed is attributed to that both hands are locked to the force plate of the handlebar of wall force plate accompanied with greater shoulder movement make arm extension changes more dramatically than two other gait conditions. Relation on great toe and trunk as well as upper body movement was also reported (Bird et al., 2003), suggesting that less symmetrical great toe pressure in RH may be affected triceps brachii muscle activation at low speed of gait as we observed in this study.

Walking holding condition resulted in greater symmetry in tibialis anterior and triceps brachii muscle activation among RH, and in soleus muscle activation among LH, at lower speed. This model of walking holding condition has been employed rehabilitation of stroke patients (Malouin et al., 1992; Laufer et al., 2001). This study supported the previous studies which also reported that such walking condition improve symmetry among stroke survivors.

On the contrary, at low speed, manual pushing generated less symmetrical pressure on great toe and 5th metatarsal among RH, indicating that maintaining balance during manual pushing is more difficult than walking with arm-swing and walking holding.

These findings also accompanied by less symmetrical triceps brachii muscle activation. Erector spinae muscle activation was also found less symmetrical compared to both walking with arm-swing and walking holding condition at 4 km/h. LH, on the other hand, did not show any significant differences due to gait condition outside soleus muscle at lower speed.

LH and RH showed different strategy in dealing with the demand of pushing force exertion during manual pushing. While LH accomplished the task by undergoing less symmetry on ankle plantar flexion at low speed, manual pushing affected RH in more areas of the body, from ankle, trunk to upper arm. At low speed, manual pushing generated asymmetrical ground pressure on great toe and 5^{th} metatarsal, and asymmetrical muscle activation of triceps brachii, while at higher speed, it generated asymmetry on erector spinae muscle activation, showing greater demand for trunk stabilisation.

6.5 Conclusions

Unlike LH, gait cycle duration and stance phase duration were more affected by gait condition in RH. Walking holding showed shorter ground contact duration on great toe and 1^{st} metatarsal in both LH and RH. Walking holding condition also resulted in greater symmetry than manual pushing and walking with arm swing. This probably confirms the importance of this condition in rehabilitation therapy of stroke survivors. The demand of force exertion in manual pushing affected LH and RH in different way. While manual pushing resulted in greater asymmetry at low speed on ankle plantar flexion on LH, it generated asymmetry to more body parts of RH depending of gait speed. At low speed, RH have asymmetry at great toe and 5^{th} metatarsal ground pressure accompanied by asymmetrical triceps brachii muscle activation, at higher speed the greater demand of trunk stability during manual pushing resulted in asymmetrical erector spinae muscle activation. The smaller number of LH probably also affected the results of our study. Future study involving more subjects based on statistical power analysis is necessary.

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Chapter 7
Conclusions

7.1 Summary of each chapter

In chapter 1, the background of this research is presented from general considerations on basic theory of symmetry, the hypothetical origins of laterality, and the association of laterality with bipedal human during evolution, some issues of laterality and adaptability to the environment, and previous studies on manual pushing. Considering that laterality during manual pushing has been given less attention compared to more common activities such as walking, this study tried to explore the characteristics of symmetry/asymmetry in manual pushing and its comparison to normal walking.

Chapter 2 describes the methods involved in this research, started from the explanation of the basic concept of biomechanics, measurement, analysis and assessment methods, followed by description of basic kinematics on body motion in anatomical plane, and kinetics based on calculation of forces generated in certain activities measured. The chapter also describes electromyogram (EMG) utility in biomechanics research such as this study, from its application to measure muscle activation, data processing, until data analysis. To measure symmetry, we employed symmetry index (SI) for static condition and cross-correlation function analysis for dynamic condition.

Chapter 3 investigated the effect of different trunk inclination on bilateral trunk muscular activity, centre of pressure, and force exertions in static pushing postures. In this study we tried to reproduce body positions during manual pushing in various static pushing postures. A proper static pushing posture should be characterized by harmony among greater pushing force exertion, acceptable muscular activity, and good balance control. The objective of this study was to explore the influence of body inclination on the activation of trunk muscles and on maximum pushing forces, vertical ground reaction forces (GRF), and the standard deviation (SD) of centre of pressure (COP) displacement to identify recommendations for good body posture in static pushing. The study found that trunk muscle activation increased along with greater body inclination. This study also indicated the existence of laterality which has been given less attention in mainstream manual pushing studies. This suggestion resulted in study on laterality in manual pushing while walking and its comparison with walking condition as investigated in chapter 4, 5, and 6.

Chapter 4 investigated the influence of laterality to different patterns of asymmetrical foot pressure and muscle activation during gait cycle in manual pushing. Seventeen subjects were made to exert pushing force while walking on a treadmill. Symmetry assessment was performed by comparing bilateral data of foot contact duration and cross-correlation function (CCF) of pressure sensors attached on footsole and EMG signals. This chapter shows that left-handers have different characteristics from right-handers during manual pushing, and that left-handers have greater asymmetry than right-handers.

Chapter 5 investigated the influence of laterality and walking speed on walking symmetry during gait. This chapter is actually presenting data from the same experiment of chapter 4 with limitation that the data analysed and discussed is only walking with arm-swing condition. The same with chapter 4, in chapter 5 also left-handers show different characteristics from right-handers based on foot contact duration, foot pressure and muscles activation. During walking, left-handers showed asymmetry in calcaneus contact duration, whereas right-handers did not show asymmetrical contact duration in all the five points of sole measured. Left-handers also showed lower cross-correlation coefficient in great toe at 1.5 and 3 km/h, accompanied by longer time lag at 3 km/h. In 1st metatarsal, at 3 km/h, left-handers showed lower cross-correlation coefficient, while in 3rd metatarsal, left-handers showed longer time lag than right-handers at 1.5 and 3 km/h. Left-handers showed longer time lag than right-handers in calcaneus at all speeds measured. In muscle activation, left-handers showed longer time lag of soleus muscle activation at 3 km/h, and lower cross-correlation coefficient of erector spinae muscle activation at 4 km/h. Thus, in general, left-handers showed greater asymmetry than right-handers during walking.

Chapter 6 compared the conditions of manual pushing, walking with arm-swing and walking holding a handlebar, with all conditions were performed on a treadmill, just like in chapter 4 and 5. Walking holding condition resulted in greater symmetry than manual pushing and walking with arm swing. This probably confirms the importance of this condition in rehabilitation therapy of stroke survivors. The demand of force exertion in manual pushing affected left-handers and right-handers in different way. While manual pushing resulted in greater asymmetry at low speed on ankle plantar flexion on left-handers, it generated asymmetry to more body parts of right-handers depending of gait speed. Different gait conditions seem affected right-handers more than left-handers, despite the latter, in all gait conditions, showed greater asymmetry than the former.

7.2 Conclusion of the study

The study on symmetry in static and dynamic manual pushing may conclude as follows:

1. It can be assumed that a greater pushing load requires a greater pushing force, which in turn is enhanced by the utilization of bodyweight through body inclination. As the body inclines lower, the step length that is required to maintain balance becomes longer; based on proximity to the centre mass of the body, greater hip flexion occurs, which in turn causes greater activation of the lower back muscles. The pushing strategy differed between below 15^o and above 15^o. While the former depended more on the stance, the latter depended more on bodyweight utilization; therefore, the larger the inclination is, the greater the pushing force became. The optimum stability during pushing was found to occur at the 30^o body inclination. A body inclination of 45^o resulted in the highest muscular activity during the staggered feet conditions, and the difference between this activation and the others was significant. Based on the pushing force exerted, stability and lower back muscular activity, the 30^o body inclination was found to be the most optimum posture for maximum pushing force exertion. Static testing was only able to represent dynamic testing at a low velocity (Resnick and Chaffin, 1995), therefore, further research on dynamic pushing, which represents the majority of daily pushing activity, is also necessary. This study also shows asymmetry in muscular activity and force exertion which have been given less attention in manual pushing studies, which is necessary to study further.
2. Gait cycle duration was found to be symmetrical in all manual pushing conditions during pushing while walking on a treadmill. Left-handers demonstrated asymmetry in calcaneus contact duration to control their ankle flexion, whereas right-handers were symmetrical. Velocity affects tibialis anterior muscle time lag and soleus muscle cross-correlation coefficients mainly in left-handers. Triceps brachii muscle cross-correlation coefficients in left-

handers were found to be affected by increasing velocity. We found that left-handers is not a mirror of right-handers, since they both have distinct characteristics, furthermore these asymmetries were not strictly associated with the preferred side, indicating that generalisation of preferred side in whole body coordination should be avoided since we cannot separate one side from the other.

3. During walking, left-handers showed greater asymmetry in contact duration and pressure timing of calcaneus, pressure of great toe and pressure timing of 3rd metatarsal at 1.5 and 3 km/h. right-handers showed that increasing walking speed from 1.5 km/h to 4 km/h improve symmetry in soleus and lumbar erector spinae muscle activation. Left-handers and right-handers did not show mirrored characteristics.
4. Walking holding condition resulted in greater symmetry than manual pushing and walking with arm swing. This probably confirms the importance of this condition in rehabilitation therapy of stroke survivors. The demand of force exertion in manual pushing affected left-handers and right-handers in different way. While manual pushing resulted in greater asymmetry at low speed on ankle plantar flexion on left-handers, it generated asymmetry to more body parts of right-handers depending of gait speed. At low speed, right-handers have asymmetry at great toe and 5th metatarsal ground pressure accompanied by asymmetrical triceps brachii muscle activation, at higher speed the greater demand of trunk stability during manual pushing resulted in asymmetrical erector spinae muscle activation.

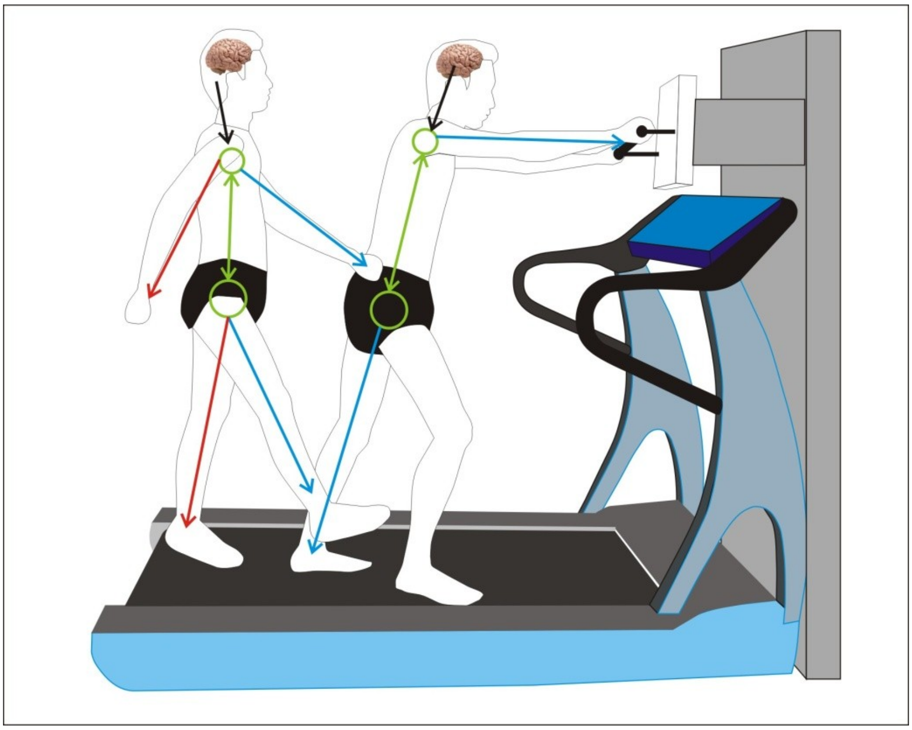

Figure 7-1. During locomotion, Dietz (2002) assumed that brain command is predominantly meadiated by interneurons, cervical and thoraco-lumbar propriospinal sysmets become coupled and coordinate arm and leg movement. Central pattern generators role during rhythmic movements such as walking has been reported in previous studies to be greater than direct cortical-motor neuronal intervention. However this assumption based on general view that asymmetry only exists in differentiated hand movements and cultural limitation imposed on them. Our findings showed difference between left-handers and right-handers in degree of symmetry during locomotion. Thus we assume that there should be certain degree of trade-off between direct cortical-motoneuronal and intraspinal mechanism during locomotion, which is different between groups of handedness.

5. In general, left-handers showed greater asymmetry than right-handers, especially in manual pushing and walking conditions. Asymmetrical walking has been associated with greater metabolic costs (Srinivasan, 2011). It is interesting to investigate this aspect furthermore in relation with lower left-handers survivability compared to their right-handers counterpart. The combination of left-handers and mixed-footers compared to right-handers and right-footers as in

our study probably had effect on less symmetrical condition among left-handers. This finding is related to the concept of cross-symmetry in interlimb coordination, where during manual pushing and walking, with regard to direction of movement, there is a contralateral combination between upper and lower limbs movement, such as when right leg perform toe-off followed by swing phase, during manual pushing, the greater pushing force would be on the left side generated by left-arm, while during walking, swing phase of the right foot would be accompanied by left arm-swing. Previous studies also reported neuronal coupling between right tibialis anterior muscle and left triceps brachii muscle (Delwaide and Crena, 1984; Dietz, 2002).

7.3 Future research consideration

Relationship between limb laterality and bilateral difference of the brain is highly interesting subject for the researchers in the field of kinesiology and sport sciences, especially in the discussion on direct motor neuronal and intraspinal mechanism role during locomotion, which has been assumed to be very important during evolution (Golubitsky et al., 1999; Dietz, 2002). Asymmetries of cortical organization have been associated with laterality (Springer and Deutsch, 1997; Soros et al., 1999; Jung et al., 2003), with less clear patterns among LH (Coren, 1992; Soros et al., 1999) which is associated with their better ability in utilizing both sides of their limbs due to environmental adaptation as well as lower interhemispheric transfer time and larger corpus callosum (Witelson, 1985, Bernard, 2011). The previous studies revealed data that left-handers have shorter reaction time than right-handers. This trait should be advantageous to left-handers, on the contrary to their lower survivability. Our research, on the contrary, found that left-handers were more asymmetrical than right-handers in manual pushing and walking, thus justify their disadvantages which probably make number of left-handers lower than right-handers during evolution, since the asymmetry demonstrated that they have less energy efficiency during locomotion, as well as less ability to react to stimuli during whole body coordination.

The different findings between our study which is based on whole body coordination than previous studies where either left or right hand of the subjects performed task in isolation from each other also probably suggest that while isolating left and right hand from each other during experiment probably simplify the procedure, it actually does not represent the actual condition accurately. A better understanding of laterality during locomotion will reveal greater potency from either group of handedness to adapt to the environment. This is very relevant to two keywords of physiological anthropology: whole-body coordination and functional potentiality. With regard to whole-body coordination, phenomenon of cross-symmetry should also be taken into consideration in the future research.

While this study shows difference between groups of handedness, we did not measure brain activities therefore we cannot analyse the relation between brain and asymmetrical locomotion in detail. Laterality itself is affected by individual difference due to cultural and social backgrounds, and in addition, laterality is not necessarily corresponds if all movements of extremities satisfy the regulation of questionnaire which was employed to assess the degree of laterality in this study. In the future, measurement of brain activities using instruments such as electroencephalogram (EEG) or functional near-infrared spectroscopy (f-NIRS) is necessary, to be able to understand better the phenomenon of brain association with laterality.

In gait therapy for patients with deficiency, such as hemiplegic patients or children who bear cerebral palsy, the method used in our study probably can be applied as a complementary to the conventional balance test. While balance test only measured statically, before and after the therapy procedure, our method is probably able to measure in real time situation dynamically. However this assumption requires further research involving a significant number of patients with longer therapy period, since our study which was performed in relatively short time with normal patients cannot justify this assumption.

7.4 References

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Hopefully, whatever I have learned during my stay in Japan will be useful for the development of scientific research when I get back to my country afterward.

APPENDIX

Waterloo Footedness Questionnaire

Instructions: Answer each of the following questions as best as you can. If you always use one foot to perform the described activity, circle Ra or La (for right always or left always). If you usually use one foot circle Ru or Lu, as appropriate. If you use both feet equally often, circle Eq. Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

1	Which foot would you use to kick a stationary ball at a target straight in front of you?	La	Lu	Eq	Ru	Ra
2	If you had to stand on one foot, which foot would it be?	La	Lu	Eq	Ru	Ra
3	Which foot would you use to smooth sand at the beach?	La	Lu	Eq	Ru	Ra
4	If you had to step up onto a chair, which foot would you place on the chair first?	La	Lu	Eq	Ru	Ra
5	Which foot would you use to stomp a fast-moving bug?	La	Lu	Eq	Ru	Ra
6	If you were to balance on one foot on a railway track, which foot would you use?	La	Lu	Eq	Ru	Ra
7	If you wanted to pick up a marble with your toes, which foot would you use?	La	Lu	Eq	Ru	Ra
8	If you had to hop on one foot, which foot would you use?	La	Lu	Eq	Ru	Ra
9	Which foot would you use to help push a shovel into the ground?	La	Lu	Eq	Ru	Ra
10	During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight at first?	La	Lu	Eq	Ru	Ra
11	Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities?	YES		NO		
12	Have you ever been given special training or encouragement to use a particular foot for certain activities?	YES		NO		
13	If you have answered YES for either question 11 or 12, please explain:					

Waterloo Handedness Questionnaire

Instructions: Please indicate your hand preference for the following activities by circling the appropriate response. If you always (i.e. 95% or more of the time) use one hand to perform the described activity, circle Ra or La (for right always or left always). If you usually (i.e. about 75% of the time) use one hand circle Ru or Lu, as appropriate. If you use both feet equally often (i.e. you use each hand about 50% of the time), circle Eq.

1	Which hand would you use to adjust the volume knob on a radio?	La	Lu	Eq	Ru	Ra
2	With which hand would you use a paintbrush to paint a wall?	La	Lu	Eq	Ru	Ra
3	With which hand would you use a spoon to eat soup?	La	Lu	Eq	Ru	Ra
4	Which hand would you use to point to something in the distance?	La	Lu	Eq	Ru	Ra
5	Which hand would you use to throw a dart	La	Lu	Eq	Ru	Ra
6	With which hand would you use the eraser on the end of a pencil?	La	Lu	Eq	Ru	Ra
7	In which hand would you hold a walking stick?	La	Lu	Eq	Ru	Ra
8	With which hand would you use an iron to iron a shirt?	La	Lu	Eq	Ru	Ra
9	Which hand would you use to draw a picture?	La	Lu	Eq	Ru	Ra

10	In which hand would you hold a mug full of coffee?	La	Lu	Eq	Ru	Ra
11	Which hand would you use to hammer a nail?	La	Lu	Eq	Ru	Ra
12	With which hand would you use the remote control for a TV?	La	Lu	Eq	Ru	Ra
13	With which hand would you use a knife to cut the bread?	La	Lu	Eq	Ru	Ra
14	With which hand would you use to turn the pages of a book?	La	Lu	Eq	Ru	Ra
15	With which hand would you use a pair of scissors to cut paper?	La	Lu	Eq	Ru	Ra
16	Which hand would you use to erase a blackboard?	La	Lu	Eq	Ru	Ra
17	With which hand would you use a pair of tweezers?	La	Lu	Eq	Ru	Ra
18	Which hand would you use to pick up a book?	La	Lu	Eq	Ru	Ra
19	Which hand would you use to carry a suitcase?	La	Lu	Eq	Ru	Ra
20	Which hand would you use to pour a cup of coffee?	La	Lu	Eq	Ru	Ra
21	With which hand would you use a computer mouse?	La	Lu	Eq	Ru	Ra
22	Which hand would you use to insert a plug into an outlet?	La	Lu	Eq	Ru	Ra
23	Which hand would you use to flip a coin?	La	Lu	Eq	Ru	Ra
24	With which hand would you use a toothbrush to brush your teeth?	La	Lu	Eq	Ru	Ra
25	Which hand would you use to throw a baseball?	La	Lu	Eq	Ru	Ra
26	Which hand would you use to turn a doorknob?	La	Lu	Eq	Ru	Ra
27	Which hand would you use for writing?	La	Lu	Eq	Ru	Ra
28	Which hand would you use to pick up a piece of paper?	La	Lu	Eq	Ru	Ra
29	Which hand would you use a hand saw?	La	Lu	Eq	Ru	Ra
30	Which hand would you use to stir a liquid with a spoon?	La	Lu	Eq	Ru	Ra
31	In which hand would you hold an open umbrella?	La	Lu	Eq	Ru	Ra
32	In which hand would you hold a needle while sewing?	La	Lu	Eq	Ru	Ra
33	Which hand would you use to strike a match?	La	Lu	Eq	Ru	Ra
34	Which hand would you use to turn on a light switch?	La	Lu	Eq	Ru	Ra
35	Which hand would you use to open a drawer?	La	Lu	Eq	Ru	Ra
36	Which hand would you use to press button on a calculator?	La	Lu	Eq	Ru	Ra
37	Is there any reason (i.e. injury) why you have changed your hand preference for any of the above activities?	YES		NO		
38	Have you ever been given special training or encouragement to use a particular hand for certain activities?	YES		NO		
39	If you have answered YES for either question 37 or 38, please explain:					

利き足アンケート

以下の各質問に答えてください。述べられている運動について、いつも右足を使う場合はRaに、いつも左足を使う場合はLaに丸をつけてください。しばしば右足を使う場合はRuに、しばしば左足を使う場合はLuに丸をつけてください。どちらの足も同じ頻度で使う場合は、Eqに丸をつけてください。

全ての質問に同じ回答をすることはせずに、順番にそれぞれの運動についてあなた自身が行うことを想像してください。必要であれば、述べられている運動の身振りをしても構いません。

	いつも左		同頻度		いつも右
1.正面にある的に向けて静止したボールを蹴るのに、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
2.片足で立たなければならないとき、どちらの足で立ちますか。	La	Lu	Eq	Ru	Ra
3.浜辺の砂を足でならすのに、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
4.椅子の上に登らなければならないとき、どちらの足を先に椅子に乗せますか。	La	Lu	Eq	Ru	Ra
5.素早く動く虫を踏みつけるのに、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
6.電車の線路の上に片足で立ちバランスをとるとしたら、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
7.つま先で小石を拾い上げようとするとき、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
8.片足で軽く跳ねるとき、どちらの足を使いますか。	La	Lu	Eq	Ru	Ra
9.ショベルを地面に差しこむときに、どちらの足でショベルを踏みますか。	La	Lu	Eq	Ru	Ra
10. リラックスして立っているとき、はじめ人は片方の足にほとんどの体重をかけ、もう片方の脚を少し曲げます。あなたは、はじめにどちらの足にほとんどの体重をかけますか。	La	Lu	Eq	Ru	Ra
11. 上記の動きのいずれかのうち、好んで使う足を変える原因となったもの（ケガなど）は有りますか。	はい	いいえ			
12. どちらか片方の足を使う特定の活動に励んだり、特別なトレーニングを受けたことがありますか。	はい	いいえ			
13. 11または12に「はい」と答えた場合は、それについて説明してください。					

利き手アンケート

以下の運動の際に、好んで使う手について該当する答えに丸をつけてください。述べられている運動の際に、いつも（95%以上の頻度）右手を使う場合はRaに、いつも左手を使う場合はLaに丸をつけてください。しばしば（75%くらいの頻度）右手を使う場合はRuに、しばしば左手を使う場合は、Luに丸をつけてください。どちらの手も同じ頻度で使う場合は、Eqに丸をつけてください。

	いつも左		同頻度		いつも右
1. ラジオの音量をつまみで調整するとき、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
2. ハケで壁を塗るのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
3. スプーンでスープを飲むのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
4. 距離の離れたものを指さすのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
5. ダーツの矢を投げるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
6. 鉛筆の端についている消しゴムを使うのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
7. 杖を持つのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
8. シャツにアイロンをかけるのに、アイロンをどちらの手で持ちますか。	La	Lu	Eq	Ru	Ra
9. 絵を描くのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
10. コーヒーがいっぱい入ったマグカップを持つのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
11. ハンマーで釘を打つのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
12. テレビのリモコンを操作するのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
13. パンをナイフで切るのに、ナイフをどちらの手で使いますか。	La	Lu	Eq	Ru	Ra
14. 本のページをめくるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
15. 紙をハサミで切るのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
16. 黒板を消すのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
17. ピンセットを使うのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
18. 本を拾い上げるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
19. スーツケースを運ぶのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
20. カップにコーヒーを注ぐのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
21. コンピューターのマウスを、どちらの手で使いますか。	La	Lu	Eq	Ru	Ra
22. プラグをコンセントにさし込むのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
23. コインをはじくのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
24. 歯を磨くとき、歯ブラシをどちらの手で使いますか。	La	Lu	Eq	Ru	Ra
25. 野球のボールを投げるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
26. ドアノブを回すのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
27. 字を書くのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
28. 紙を拾うのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
29. のこぎりをどちらの手で使いますか。	La	Lu	Eq	Ru	Ra
30. 液体をスプーンでかき混ぜるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
31. 傘をさすとき、どちらの手で傘を持ちますか。	La	Lu	Eq	Ru	Ra
32. 縫い物をするとき、どちらの手で針を持ちますか。	La	Lu	Eq	Ru	Ra
33. マッチを擦るのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
34. 照明のスイッチを入れるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
35. 引き出しを開けるのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
36. 電卓のボタンを押すのに、どちらの手を使いますか。	La	Lu	Eq	Ru	Ra
37. 上記の動きのいずれかのうち、好んで使う手を変える原因となったもの（ケガなど）は有りますか。		はい	いいえ		
38. どちらか片方の手を使う特定の活動に励んだり、特別なトレーニングを受けたことがありますか。		はい	いいえ		
39. 37または38に「はい」と答えた場合は、それについて説明してください。					