Mechanical Energy Exchange during Plane Walking and Stepping over Obstacle in Subjects with Down Syndrome

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to my beloved parents
and my brothers, who are my best friends,
Amirtaher and Amirtayeb

“Seek knowledge
from the cradle to the grave”

Muhammad [pbuh]

Abstract

The purpose of this study is to compare mechanical energy exchange, and mechanical work and power generated or absorbed at the lower limbs joints in young adults with Down syndrome (DS) (21.6±7 years) with an age-matched control group of healthy subjects (CG, N) (25.1±2.4 years). The subjects walked along a walkway in two conditions: plane walking, and stepping over an obstacle (10% of the subject's height). The tasks were acquired using an optoelectronic system for quantitative movement analysis. Also ground reaction forces were sampled from two force plates mounted in the middle of the walkway. Spatiotemporal, mechanical energy, and joints' power and work parameters have been obtained and analyzed respectively with Smart Analyzer (Elit2002, BTS) and proper statistical software. Spatiotemporal parameters demonstrated a different motor strategy in DS compared to N in both conditions. DS walked slower, with shorter step length, and greater step width in both conditions. While potential energy exchanges were similar between groups, kinetic energy was lower in DS compared to N in both conditions. However, according to the literature, DS, compared to N used different motor strategies; they had similar values of mechanical energy recovery (ER) in all conditions. It means that DS can recover almost the same percentage of consumed mechanical energy with N during a gait cycle. Also, results showed that powers generated or absorbed in lower limbs joints and work produced in these joints were different across groups.

Keywords: gait, mechanical energy, power, work, Down syndrome, obstacle

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List of Abbreviations

IQ	Intelligence quotient
DS	Down syndrome
N	normal
CG	control group
wlk	plane walking
obs	stepping over obstacle
CM	center of mass of the body
SSE	sum of segmental energies
EE	energy expenditure
MET	metabolic equivalent of task
PE	potential energy
KE	kinetic energy
TME	total mechanical energy
ER	mechanical energy recovery
P	net joint power
W	work generated or absorbed at the joints
ANOVA	analysis of variance
AP	antero-posterior direction
ML	medio-lateral direction

INTRODUCTION

Gait analysis is the study of human locomotion, and is usually used for measuring body movements, body mechanics, the activity of the muscles, and motor system strategies. It is applied to analyze the walking ability. Therefore, gait analysis is considered as a tool for treatment of individuals with walking abnormality or walking disability. Gait can be defined as a cyclic event, which simultaneously propels the body forward and maintains stance stability. As an individual advances, one limb provides a base of support, while the other limb is shifted forward to create a new base. This ability can be severely impacted by physical and/or cognitive impairment causing weakness or loss of function, as in the case of Down syndrome (DS).

As gait analysis, by itself, cannot reveal all the aspects and characteristics of human's motor system, a growing interest was observed for studying more complex movements, such as running, jumping, stepping over obstacle, and climbing the stairs during last years. So, different tests have been suggested to use these complex movements in order to help medical diagnosis. Particularly, analysis of stepping over obstacle in healthy subjects was object of different works in recent years. On the contrary, there are few studies in this area (obstacle avoidance) in subjects with pathological conditions such as DS. The observations show that subjects with DS usually have difficulty when they encounter with the obstacle. Therefore, this study was concentrated on the obstacle avoidance in subjects with DS. All of existing studies on obstacle avoidance in subjects with DS are about the kinematic aspects of gait; and according to the best of our knowledge, there is no study about calculation of mechanical energy in plane walking and stepping over obstacle in subjects with DS. A mechanical energy comparison between healthy subjects and subjects with DS can reveal whether they use different energy strategies during plane walking and stepping over obstacle or not. For these reasons, the main goal of the study was to provide an estimation of mechanical energy exchanges in subjects with Down syndrome when they are stepping over an obstacle.

In this study, some of the spatiotemporal, mechanical energy, power, and mechanical work parameters of a group of subjects with DS were analyzed and compared with those of normally developed subjects during plane walking and stepping over obstacle. The present thesis is divided into five chapters. In the first chapter, the main characteristics and features of DS are explained. Then, there is a literature review on "gait analysis in subjects with DS". In this part, main characteristics of DS's plane walking will be discussed. In the last part of the chapter, a review on the literature related to the "obstacle avoidance in subjects with DS" will be provided. In the first part of the second chapter, there is a review on the previous studies related to the mechanical energy exchange and work during plane walking and stepping over obstacle in healthy subjects and subjects with DS. Then, different "computational methods" for the assessment of energy expenditure (EE) are presented. The chapter will be concluded with a comparison between above mentioned methods and their application. The third chapter is about "materials and methods" that were used in the study. This chapter is divided into four parts: "participants", "acquisition and instrumentation", "parameters", and "statistical analysis". Finally, results of the study are presented and discussed in the last two chapters.

CHAPTER ONE • DOWN SYNDROME AND STUDIES ON MOVEMENT ANALYSIS

Down Syndrome (DS) is the most common chromosome abnormality in humans. It is a genetic condition which in most cases is due to presence of a third copy of chromosome 21. In other words, a person with Down syndrome has 47 chromosomes instead of the usual 46. This usually happens when the individual inherits two copies of chromosome 21 (instead of one) from the mother's egg during fertilization. In rare cases, the individual inherits the extra chromosome 21 through the father's sperm.

1.1 Main Characteristics and Features

A syndrome is defined as a special combination of characteristics that are consistently found together. Frequently this is associated with a change in genetics. DS was discovered by finding consistent features and physical characteristics that were consistently together along with a mental disability (mental retardation or MR). The average intelligence quotient (IQ) of young adults with DS is about 50 (Liptak, 2008), while healthy young adults typically have an IQ of 100 (American academy of pediatrics, 2001). Beside mental retardation, people with DS share a number of physical characteristics. Not everyone will have all of them, but they may include:

- Reduced muscle tone which results in floppiness (hypotonia)
- Flexible ligaments
- A big space between the first and second toe (sandal gap)
- Broad hands with short fingers
- Their palm may have only one crease across it (palmar crease)
- A small nose and flat nasal bridge
- A small mouth
- Eyes that slant upwards and outwards
- A below-average weight and length at birth

A certain degree of developmental disability is always seen in individuals with Down syndrome, as far as their mental capabilities are concerned. This, however, does not mean that it is impossible for them to learn or process information. However, the rate of learning and information processing is slower in them, as compared to unaffected people. This, to a great degree, can be corrected by early intervention involving appropriate teaching methods, lots of motivation and positive reinforcement, and pushing them constantly towards improvement while getting them access to good education. A common cognitive characteristic of people with Down syndrome is their ability to understand more than they can express. More often than not, these individuals have trouble expressing what they have learned or understood via the conventional outlets of speech and writing. To overcome this constraint, they should be encouraged to use other means of expressions such as pictures, colors, sounds, or any other media, to express. Often, this inability to properly express themselves is misunderstood as an indication of learning disability.

Along with these physical characteristics, children with Down syndrome suffer from a wide range of health problems associated with this disorder. Some of the most common medical characteristics accompanying this syndrome are:

- Congenital heart defects
- Gastrointestinal disorders
- Respiratory problems
- Childhood leukemia (somewhat rare, but possible)
- Increased susceptibility to infections

Cardiac abnormalities are one of the most serious medical issues faced by individuals with Down syndrome. People with Down syndrome suffer from progressive heart problems. Atrioventricular septal defect is the most common congenital heart defect. Medications and lifelong cardiac screening are common in children with Down syndrome. In some cases, a heart surgery is indicated.

People with Down syndrome have a depressed immune system. As a result, they are at a higher risk of infections. Some respiratory problems such as frequent cold, cough, and flu are common in these people (Mao et al, 2005).

Other medical problems associated with Down syndrome are vision and hearing problems, epilepsy, skin problems, gastrointestinal disorders and thyroid issues. People with Down syndrome often have a strong tendency towards obesity. They may suffer from hypothyroidism, hyperthyroidism and a shortage of growth hormones. Infants with Down syndrome have very soft skin. As they grow older, their skin becomes coarse and dry. Atopic dermatitis or atopic eczema is the main skin problem found in children with

Down syndrome. They also suffer from some gastrointestinal disorders including the anatomical abnormalities such as annular pancreas, aganglionic megacolon, imperforate anus, and functional disorders such as esophageal motility disorders, gastroesophageal reflux, and malabsorption. These people are at a greater risk of developing Alzheimer's disease.

As mentioned above, all children with Down syndrome have some degree of learning disability and delayed development but this varies widely between individual children. Babies with Down syndrome also often have short arms and legs and low muscle tone making it harder for them to learn how to move. Certain development milestones are often affected, including: reaching, sitting, standing, walking, communicating, talking, and reading.

1.2 Gait Analysis in Subjects with DS

Most of the literature concerning movement analysis in DS has focused on the evaluation of plane walking.

Locomotion is a fundamental element of everyday life. Gait can be defined as a cyclic event, which simultaneously propels the body forward and maintains stance stability. As an individual advances, one limb provides a base of support, while the other limb is shifted forward to create a new base. This ability can be severely impacted by physical and/or cognitive impairment causing weakness or loss of function, as in the case of DS. Some children with DS will begin walking at around 2 years of age, while others will not walk until age four.

Rigoldi et al (2010) studied the effects of aging in a group of subjects with DS and compared them with an age-matched group of healthy subjects. The study was performed on 32 individuals with DS and 36 healthy subjects (10 children, 15 teenagers, and 16 adults). They reported that the participants with DS develop a strategy focused on the reduction of the degrees of freedom, increasing the dispersion of generated power in the frontal plane; while in healthy individuals the strategy is focused on the use of all the degrees of freedom, in order to reach the effectiveness of the gesture and finalize their movement in sagittal progression (Rigoldi et al, 2010). In other words, while sagittal plane was the main plane of movement in normal subjects, DS followed a different walking strategy. Also, they reported that N subjects use all joints in a specific way, all joints are involved with a specific task in order to obtain a synergic movement; instead, DS diminish the number of body segments involved in the normal gait using co-contraction and consequently increasing joint stiffness. In that way, they can better control the limited number of body segments. In addition, according to literature, children and teenagers with DS walk with lower values of step length and velocity compared to normal age-matched group. By aging, DS subjects increase step length value, reaching a better balance control that allows them to walk with step length that is closer to normal data (Smith & Ulrich, 2008; Rigoldi et al, 2010; Vimercati et al, 2011). Along with lower values of step length, Vimercati et al (2013) reported that DS represented lower maximum foot elevation during plane walking.

There are characteristic changes of gait pattern in people with DS. Increased hip flexion and increased knee flexion at initial contact, in DS compared to healthy group, are mentioned in the literature (Galli et al, 2008). In literature, joint stiffness is defined as the resistance that the joint offers during gait in response to an applied moment (Davis & De Luca, 1996). Galli et al (2008) reported increased hip joint stiffness and decreased ankle joint stiffness in DS compared to healthy subjects.

Also there are several studies about fall risk in subjects with DS. According to the literature, aging is associated with fall risk. Generally, increased spatial and temporal variability is associated with increased fall risk (Maki, 1997). Increased movement patterns (e.g. stride velocity variability) in persons with DS compared to their healthy peers, and increased step width variability in older as compared to younger adults are reported in the literature (Kubo & Ulrich, 2006; Maki, 1997; Owings & Grabiner 2004).

1.3 Obstacle Avoidance in Subjects with DS

Successful interaction with the environment requires the adaptation and combination of fundamental locomotor skills; this ability to combine movement is essential to daily living. One example of an everyday skill is stepping and walking over an obstacle. In order to do this individuals must process sensory and movement related information and coordinate a motor response in a challenging environment. Whilst the principles of obstacle crossing are similar to most of normal gait, obstacles present a significant hazard to the population, and so it has been taken into consideration by the literature.

Several studies analyzed the gait parameters during stepping over obstacle in healthy subjects. In healthy subjects, obstacle avoidance is directed in antero-posterior direction, with a flex-extension joint movement. In addition healthy subjects follow conservation strategy during their movement. Conservation strategy is defined as maintaining most of the walking parameters unvaried (Vimercati et al, 2011). In stepping over obstacles subjects increased obstacle-crossing step lengths and reduced obstacle-crossing speed as a function of obstacle height (Begg et al, 1998). In another study, Chen et al observed that subjects increased their average step length from 673 to 738 mm when stepping across a zero-height obstacle in the form of tape stuck to the floor (Chen et al, 1991). Chen et al concluded that subjects may have adapted their gait to accommodate the constraint due to the presence of the obstacles.

Different strategies are reported for healthy younger and older adults. Chen et al (1991) and Weerdensteyn et al (2005) reported a step shortening strategy in older adults in presence of an obstacle, while younger adults maintain most of their walking parameters unvaried and follow a conservatism strategy (with slower crossing speed, shorter step length, and shorter obstacle-heel strike distance). They interpreted the elderly's behavior as a safety strategy or as a difficulty in interpreting the sensory input given by the obstacle (Chen et al, 1991; Weerdensteyn et al, 2005).

Chou et al (1997) measured the trajectories of the swing ankle during level walking and stepping over obstacles of different heights, and compared them with predicted trajectories that were based on the criterion of minimum mechanical energy for eight healthy young adults (when stepping over obstacles, the predicted trajectories of the swing ankle were just high enough for the swing toe to clear the obstacles). They found that toe clearance increase in presence of an obstacle compared to plane walking, but remain roughly unvaried over different heights of obstacle (Chou et al, 1997).

Some studies have focused on the differences between leading and trailing limbs in crossing over obstacle. According to the literature, the limb that clears the obstacle first is defined as the leading limb and the limb that follows the leading limb is the trailing limb (the support limb during the leading limb clearance and clears the obstacle after it). Begg et al (1998) found higher toe clearance for the trailing limb than the leading limb. In addition, the trailing foot pattern was approximately symmetrical and narrow, whereas leading foot raised steeply before a gradual descent, making the profile skewed (Begg et al, 1998). In another study, Patla et al (1996) found that while toe clearances when clearing smaller obstacles are lower, there is not consistent correlation between the toe clearance values of the leading and trailing limbs. The variability in toe clearance was higher for the trailing limb, which is attributable to lack of visual exproprioceptive input about trailing limb movements and to the shorter time available following toe-off to fine-tune the trailing limb trajectory. Also they found different strategies for stepping over obstacles made by different materials (solid and fragile) with the same dimensions. Subjects walked over the fragile obstacle with higher toe clearance. They suggested that in addition to visually observable properties of obstacles such as height or width, other properties, such as rigidity or fragility, which may be classified as visually inferred, also influence the limb trajectory (Patla et al, 1996).

While a great number of studies have been performed about walking over obstacle in healthy subjects, there is a lack of investigation on subjects with DS in this area. Among the few numbers of studies, Virji-Babul and Brown (2004) studied about the mechanism of anticipatory control of gait in relation to the perception of an obstacle. Six typically developing children (range: 4-7 years old) and five children with DS (range: 5-6 years old) participated in their study. The study was performed in two different conditions: stepping over a subtle obstacle that was placed at a very low distance from the floor (1% of total body height), and an obvious obstacle that was placed at a much higher distance from the floor (15% of total body height). Virji-Babul and Brown found that subjects with DS are able to extract information about obstacle height and match this information to their movement. Usually they maintained their typical gait patterns and waited until they reached the obstacle to extract the visual information (Virji-Babul & Brown, 2004).

While normal subjects maintain a similar foot excursion between walking and stepping over obstacle, subjects with DS increase their maximum height during stepping over obstacle, reaching normal values. Normal subjects maintain unvaried values of toe clearance in both conditions. During plane walking, maximum foot elevation is lower in DS compared to Normal subjects. This can be a cause of increased tipping in DS. DS subjects increase their hip and knee flexion, and increase dorsi-flexion of the ankle to avoid the obstacle. Thus, on the sagittal plane a strategy similar to normal subjects is found, even though the results of flexion were higher for DS. In the other planes, pelvic obliquity and rotation increase together with hip abduction, with values higher than normal subjects (Vimercati et al, 2011).

Also, Smith and Ulrich (2008) examined the gait patterns of older adults with DS for precocious stabilizing adaptations during comfortable over-ground walking and in more challenging conditions. Twelve adults with DS and twelve with typical development between ages 35 and 62 years participated in the study. Participants walked barefoot in three different conditions; plane walking, stepping over obstacle (placed 12 cm above and perpendicular to the walkway), and stepping up onto and stepping down from a standard step (20 cm height, 91 cm wide, 28 cm deep, made of wood). Results showed that older adults with DS demonstrated precocious stability-enhancing adaptations in gait. To achieve this increased stability, they adapted in ways seen across the lifespan in those with DS (e.g. wider step width) and in ways used by healthy elderly adults (e.g. shorter stride lengths, slower speed, more time in stance and double support). These changes take place, however, at a much younger chronological age in adults with DS compared to their healthy peers (Smith & Ulrich, 2008).

CHAPTER TWO • STATE OF THE ART ON ENERGY ASSESSMENT

Energy expenditure and mechanical work produced at the body joints during walking are considered as two noteworthy subjects in the literature (Sutherland, 2005), because energy expenditure measurements may provide an informative data for assessing walking efficiency. In addition, mechanical work computation is important because it is an indicator of energy generation or absorption in muscles and the body system.

In the first subchapter, the literature relevant to mechanical energy exchanges is reviewed. The main computational methods of energy assessment are discussed in the second subchapter. In the final part is talked about the applications of the methods that are brought up in the second subchapter, and a brief comparison between them.

2.1 Mechanical Energy Exchange and Work

2.1.1 Mechanical Energy Exchange during Walking

Three extrinsic factors are related to gait: the self-selected gait speed adopted by the person, the smoothness of the displacement of the center of the body mass (CM), and the efficiency of the pendulum-like mechanism of walking. This mechanism of walking can be compared to the one of an inverted pendulum. At each stride, the CM is successively behind, or in front of the point of contact with the foot on the ground. When the CM is behind the point of contact, the link to the ground causes a forward deceleration, and therefore a decrease in kinetic energy, and a vertical rise in the CM, and therefore an increase in gravitational potential energy. Some of the kinetic energy due to the forward speed is converted into gravitational potential energy. As the CM moves forward of the point of contact on the ground, the link to the ground allows a decrease in the height of the CM and a concomitant increase in the forward speed, as some of the gravitational potential energy is converted back into the kinetic energy like a pendulum. This occurs twice in each gait cycle (one for the right foot, and one for the left foot). In the other words, during each gait cycle, potential energy and kinetic energy of the CM oscillate between two maximum and two minimum values. With each cycle, some of the kinetic energy is converted into potential energy and vice versa. During walking, some muscles work to increase potential and kinetic energies (positive work) and some others work to decrease it (negative work). But both these two groups work together to sustain the mechanical energy changes of the center of mass. During walking both the positive and negative works are reduced by the pendular transduction of potential energy to kinetic energy and vice versa (Cavagna et al, 1963). The fraction of mechanical energy recovered due to this transduction defined as the recovery of mechanical energy (Cavagna et al, 1976).

In normal gait, the energy cost, expressed in J kg^{-1} m^{-1}, depends mainly on gait speed and reaches a minimum at a speed which is defined as optimum, while increases progressively at speeds that are either higher or lower. In normal gait, the optimum speed is about 4-5 km h^{-1}. Generally, subjects with different motor disabilities cannot attain this speed; thus, an increase of cost of gait might well be due partly to the low speed itself. Moreover, it has been hypothesized that the increase in energy cost could be also related to abnormal kinematics of the lower limbs that disturb the smoothness sinusoidal displacement of the CM, increasing the mechanical work done to move the CM and disturbing the efficiency of the pendulum-like mechanism (Tesio et al, 1991).

Some studies are focused on energetic cost associated with body CM vertical movement (in able-bodied subjects). The CM rises in each step as the person vaults over a single extended support limb. One might then conclude that the energy requirements of gait could be reduced by actively decreasing or minimizing vertical CM movement. However recent studies show that humans do not minimize vertical CM displacement during gait (Ortega & Farley, 2005). Gordon et al showed that reduced CM displacement is not advantage for either metabolic energy economy or the reduction of mechanical work at the joints, and energy expenditure (EE) increased even though subjects walked with shorter strides to reach a lower CM vertical movement (Gordon et al, 2009).

2.1.2 Mechanical Energy Exchange during Walking in DS

Individuals with DS have joint laxity, muscle hypotonia, reduced strength, and deficits at the central nervous system (CNS) collectively considered responsible for their lower stability during gait. People with DS appear to compensate the reduced stability by walking at given speeds with higher step frequencies and greater step width adjustments manifested in greater step width variability. Agiovlasitis et al (2009) suggested a gait pattern with lesser stability and greater energetic cost among adults with DS, particularly at fast speeds. The differences in CM motion and stepping behaviors exhibited by adults with DS was one of the reasons that why these individuals showed greater energetic cost during walking respect to adults without DS. The mean vertical CM displacement did not differ between DS and healthy subjects at speeds below Froude number[1] of 0.6, suggesting a similar passive conversion between potential and kinetic energy. However the decrease in vertical CM motion at 0.6 among adults with DS is indicative of a reorganization of the movement pattern in anticipation of a transition to running (Farley & Ferris, 1988), which occurs at a slower speed in adults with DS and is associated with increased energetic cost (Agiovlasitis et al, 2009).

2.1.3 Mechanical Energy Exchange during Obstacle Avoidance

For healthy people, Chou et al reported that when stepping over obstacles, gait is not energy efficient and that conservation of energy becomes a less dominant criterion for governing the motion of the swing limb than when walking on the level ground. On the other hand, safety may become a more dominant criterion than energy cost when stepping over an obstacle. Safety strategy is defined as step-shortening strategy (Chou et al, 1997).

2.1.4 Mechanical Work

Chen et al (1997) investigated the influence of walking speed on mechanical work during gait. They reported that the work generated at the knee and hip joints were

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[1] Froude number: a dimensionless number defined as the ratio of a characteristic velocity to a gravitational wave velocity. It is defined as:

$$Fr = \frac{v}{c}$$

where v is a characteristic velocity, and c is a characteristic water wave propagation velocity.

sensitive to velocity changes, but the joint power at the ankle remained constant despite increasing speed. Teixeira-Salmela et al (2008) studied the relation between cadence and mechanical work. They showed that power, mechanical work, and the contributions of individual joints to the total energy generated and absorbed were influenced by the walking cadence.

When stepping over the obstacle, a significantly larger amount of work than that of plane walking was required. This larger amount of work not only elevated the foot to clear the obstacle, but also appeared to produce a large safety margin between the toe and the obstacle. Also, as obstacle height increased, the required mechanical work increased. This increment in mechanical work is due to support the body for a larger period of stance time as indicated by the larger period of swing time of the swing limb. The authors showed the task for different heights of the obstacle. The total amount of work required to generate the measured and predicted trajectories became less different as the height of the obstacle increased (Chou et al, 1997).

2.2 Computational Methods

Human locomotion involves smooth advancement of the body through space with the least mechanical and physiological energy expenditure. While the goal of walking is progression in the forward direction, limb motion is based on the need to maintain a symmetrical, low amplitude displacement of the center of gravity of the head, arms, and trunk in the vertical and lateral directions. This conserves both kinetic and potential energy and is the principle of biological 'conservation of energy' (Waters & Mulroy, 1999).

In pathological gait, as in individuals with DS, deviations in gait pattern can lead to inefficient gait. Gait efficiency is usually quantified by the assessment of energy expenditure (EE) during walking, which can be defined by different methods. There are two main methods for this purpose:

- calculation of metabolic energy expenditure
- calculation of mechanical energy fluctuation

Metabolic EE, which accounts for all possible sources of EE, is often seen as gold standard. However, it can only be assessed over a steady state period of walking, and therefore only detect overall EE without any specific information on causes of increased EE. Mechanical energy exchange estimation instead can provide valuable additional information in pathological gait, as it can be plotted over the gait cycle. Furthermore, mechanical energy exchange estimation does not need a steady state period and can therefor also be assessed in patients for whom walking is no longer an aerobic task (Van de Walle et al, 2011).

2.2.1 Metabolic Energy Expenditure Assessment

In this method, O_2-consumption (VO_2) is measured breath by breath, and then EE is derived based on information on VO_2 and estimated metabolic events in the body.

The ‘Metabolic Equivalent of Task’ (MET), or simply ‘metabolic equivalent’, is a physiological measure expressing the energy cost of physical activities and is defined as the ratio of metabolic rate (and therefor the rate of energy consumption) during a specific physical activity to a reference metabolic rate, set by convention to 3.5 ml $O_2.kg^{-1}.min^{-1}$ or equivalently:

$$1\,MET \equiv 1\frac{kcal}{kg * h} \equiv 4.184\frac{kJ}{kg * h}$$

Originally, 1 MET was considered as the 'Resting Metabolic Rate (RMR)' obtained during quiet sitting (Ainsworth et al, 1993). The range of MET values of different activities is from 0.9 (sleeping) to 23 (running at 22.5 km/h).

In one of the metabolic energy calculation models, EE is calculated on the basis of information on VO_2 (or METs), respiratory quotient (RQ) and caloric equivalent. RQ describes the ratio between carbon dioxide produced and oxygen consumed in metabolism, varying from 0.70 to 1.00. RQ has a well-established deterministic relationship with caloric equivalent (=thermal equivalent), which describes the amount of energy expended per one liter of consumed oxygen, varying from 4.69 to 5.05 kcal/l O_2 (McArdle et al, 1996)

2.2.2 Mechanical Energy Estimation

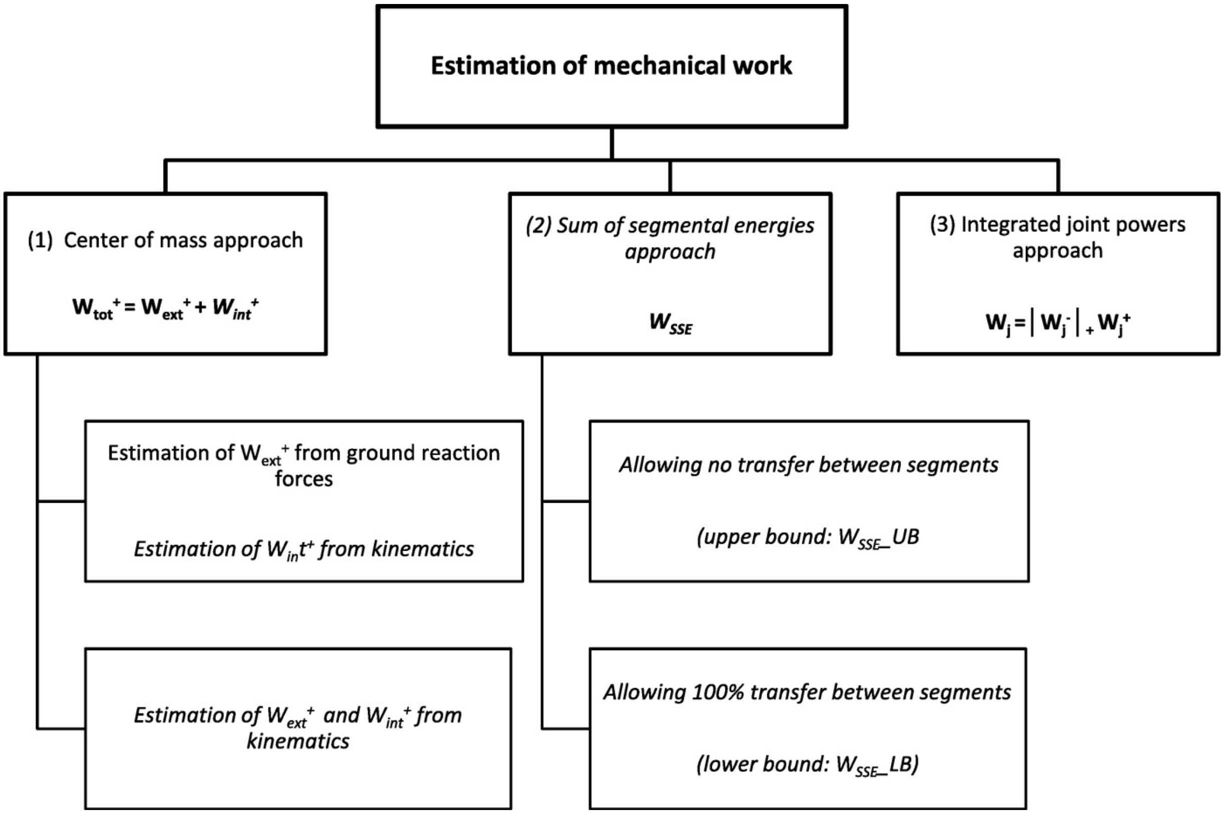
Three following approaches for mechanical energy estimation are present in the literature:

- Center of Mass approach (CM)
- Sum of Segmental Energies approach (SSE)
- Integrated joint power approach

Among the approaches mentioned above, the CM-approach is most frequently used in the literature.

In Figure 1, above mentioned approaches used for estimation of mechanical energy are listed. Each of these approaches is based on different biomechanical assumptions. Depending on what data are collected (kinematic or kinetic data) and how they are processed (allowing transfer or not between segments), protocol can differ within an approach. In the following figure, approaches and protocols in italics use kinematic data as the start point, all other approaches and protocols start from kinetics.

Figure 1 - Approaches used for mechanical energy estimation (Van de Walle et al, 2011). Mechanical energy approaches can be categorized based on different biomedical assumptions (e.g. the approach starts from kinematic or kinetic data). Here approaches and protocols in italics use kinematic data as the start point, all other approaches and protocols start from kinematics.



The parameters that are used in Figure 1 are explained in Table 1.

Table 1 - The explanation of parameters that are involved in the approaches used for mechanical energy estimation

	Explanation
W_{ext}^{+}	positive work needed to move the CM relative to the surrondings
W_{int}^{+}	positive work of the segments related to the CM
W_{tot}^{+}	positive work as estimated from CM approach
W_{SSE}	work estimated by sum of segmental energies approach (contains both positive and

	negative work)
W_{SSE}_UB	upper bound of work by W_{SSE}, no transfer allowed between segments
W_{SSE}_LB	lower bound of work by W_{SSE}, allowing all possible transfer between segments
W_j^-	negative work estimated by the integrated joint powers approach
W_j^+	positive work estimated by the integrated joint powers approach
W_j	work estimated by the integrated joint powers approach

2.2.2.1 Center of Mass Approach (CM)

This method allows the analysis of energy changes of the center of mass of the body relative to the surroundings (positive external work, W_{ext}^+) and of the body segments relative to the CM (positive internal work, W_{int}^+) where total work (W_{tot}) is $W_{ext}^+ + W_{int}^+$. From a mechanical point of view, combined movements of different body segments during locomotion are the result of the interaction between muscle activity, dictated by the central nervous system, and the mechanical demands of the locomotor activity. The motion of the center of mass of the body, representing the whole body system in movement, is the ultimate result of both energy expenditure and motions of the body segments. The work done by muscles to translate the CM (external work) with respect to the ground is one determinant of the energy expenditure of gait (Cavagna et al, 1963). W_{ext}^+ can be computed according to Cavagna's formulation:

$$E_{pot} = M_{tot} * 9.81 * CM(z)$$

$$E_{kin} = \frac{1}{2} * M_{tot} * \left(v_x^2 + v_y^2 + v_z^2\right)$$

$$E_{tot} = E_{pot} + E_{kin}$$

$$W_{ext}^+ = \sum_{i=1}^{GC} \Delta E_{tot}$$

while M_{tot} is the total body mass, GC is the number of instants during the gait cycle, and CM(z) is the vertical displacement of CM (Van de Walle et al, 2011).

2.2.2.2 Sum of Segmental Energies approach (SSE)

This method shows the analysis of energy changes of moving body segments (sum of segmental energies, W_{SSE}). Winter introduced a new definition for calculation of mechanical energy expenditure, which not only accounts for any external work but also for the internal work done by the limbs themselves. W_{SSE} should be calculated without any inter-segmental energy transfer (upper bound; W_{SSE}_UB), as well as with all possible transfer (lower bound; W_{SSE}_LB). W_{SSE}_UB can be calculated by determining total energy per segment ($E_{segment}$) and summing the increments in $E_{segment}$ over the gait cycle ($W_{segment}$). W_{SSE}_UB equals then the sum of $W_{segment}$ of all the segments, thus allowing no transfer between them. W_{SSE}_LB can be calculated summing all segmental energies ($E_{segment}$) at each instant of the gait cycle (E_{tot}_instant), thus allowing transfer between segments. W_{SSE}_LB equals then the sum of increments in E_{tot}_instant over the gait cycle (Winter, 1979; Van de Walle et al, 2011):

$$W_{SSE-UB} = \sum_{i=1}^{N} \left|\Delta E_{segment}\right|$$

$$E_{segment} = E_{pot} + E_{kin}$$

while N is the number of segments.

$$W_{SSE-LB} = \sum_{i=1}^{GC} |\Delta E_{tot-instant}|$$

$$E_{tot-instant} = \sum_{i=1}^{N} E_{segment}$$

2.2.2.3 Integrated Joint Power approach

This method represents the integration of power around the joints (net joint work, W_j). W_j can be obtained by separate integration of positive and negative net joint power profiles for neck, shoulders, elbows, wrists, waist, hips, knees, and ankles. Positive and negative work should be separately summed. The sum of positive and negative net joint work of all joints, respectively, give positive and negative net joint work of the whole body (W_j^+ and W_j^-). W_j equals then the sum of W_j^+ and $|W_j^-|$ (Winter, 2005; Van de Walle et al, 2011):

$$W_j^+ = \sum_{i=1}^{N} \sum_{i=1}^{GC} \Delta P_{joint} . \Delta t \, , \qquad if \ P_{joint} > 0$$

$$W_j^- = \sum_{i=1}^{N} \sum_{i=1}^{GC} \Delta P_{joint} . \Delta t \, , \qquad if \ P_{joint} < 0$$

$$W_j = W_j^+ + |W_j^-|$$

where GC is the number of instants during the gait cycle, and N is the number of segments.

2.2.2.4 A comparison between CM, SSE, and Integrated joint power approaches

However metabolic EE is the best method to take into account all possible sources of EE, it has some problems compared to mechanical energy exchange approaches. First of all, metabolic EE must be assessed over a steady state period of walking. As mechanical energy approaches do not have this limitation, they can easily use in those patients for whom walking is no longer an aerobic task.

For clinical use, parameters should be sensitive enough to reveal the differences between control groups and groups with pathological conditions. Among the approaches, integrated joint power approach has the best sensitivity. Moreover, this approach fits very well with metabolic EE results. On the other hand, while CM approach underestimates total EE, SSE approach overestimates it (Van de Walle et al, 2011).

2.3 Application of the Three Methods

According to the literature, potential energy is the component of a body's mechanical energy associated with its position relative to other bodies (including the ground). For example, gravitation potential energy of a body is proportional to the height of the CM above the ground. Potential energy can also be developed through stretch of a muscle and this type of potential energy is sometimes referred to as elastic potential. Potential energy is sometimes also referred to as stored energy. Similarly, kinetic energy is the component of a body's mechanical energy that is due to its motion. When a body is at rest, the kinetic energy is zero and kinetic energy reaches a maximum at maximum velocity. In the same row, mechanical energy is introduced in the literature as the energy state (potential and kinetic) of any limb segments or total body system at an instant in time. Mechanical energy recovery, the parameter which was defined by Cavagna (1976), is an indicator of how much of the mechanical energy can be recovered due to conversion of potential energy to kinetic energy and vice versa (according to pendulum model).

Mechanical power is the indicator of the rate of change of mechanical energy (also the rate of doing work) at an instant in time. Mechanical power can also be calculated as the dot product of a moment and an angular displacement. The work done on a system is equal to the change in mechanical energy in a system (segment or total body) over that same period of time. The area under the power curve provided the mechanical work generated or absorbed at the joints.

CHAPTER THREE • MATERIALS AND METHODS

As it was discussed in the first chapter, there are great numbers of studies which are about gait analysis in healthy subjects during walking and stepping over obstacle. In addition, lots of researchers have focused on gait analysis in subjects with some pathological conditions such as Down syndrome, and compared them with normally developed subjects to find the differences between gait parameters of these two groups (if any).

Also, as mentioned in chapter two, different approaches for estimation of mechanical energy exchanges were widely used for assessing walking efficiency in healthy subjects, as well as in subjects with some mental or physical disabilities.

Although there are many studies about obstacle avoidance in healthy subjects, there are few studies in this area in subjects with pathological conditions such as DS. All of these studies concerned the kinematic aspects of gait; and according to the best of our knowledge, there is no study about calculation of mechanical energy in plane walking and stepping over obstacle in subjects with DS. For these reasons, the aim of the study was to provide an estimation of mechanical energy exchanges in subjects with Down syndrome when they are stepping over an obstacle.

3.1 Participants

A total of 39 individuals were included in our study, 21 subjects with DS (age: 21.6±7 years) and one control group (CG) of 18 normal subjects (age: 25.1±2.4 years) in order to compare with DS data. Mean age, height, and weight were obtained for each group (Table 2). The data of subjects with DS were collected in the Posture and Motion Analysis Lab of San Raffaele Pisana IRCCS, TOSINVEST Sanità, Rome, whereas CG data collection was performed in Posture and Movement Analysis Laboratory, Luigi-Divieti, at Politecnico di Milano, Milan. All participants were instructed to walk at a comfortable speed. The subjects and their legal tutors gave their informed consent to the study. Inclusion criteria for the DS group were young adult age, low to medium intelligence quotient (IQ), no severe obesity, no clinical sign of dementia, and no orthopedic problems. Inclusion criteria for N subjects were the same with those for DS group, except for IQ. For normal subjects, IQ was not measured and it was considered in the normal range.

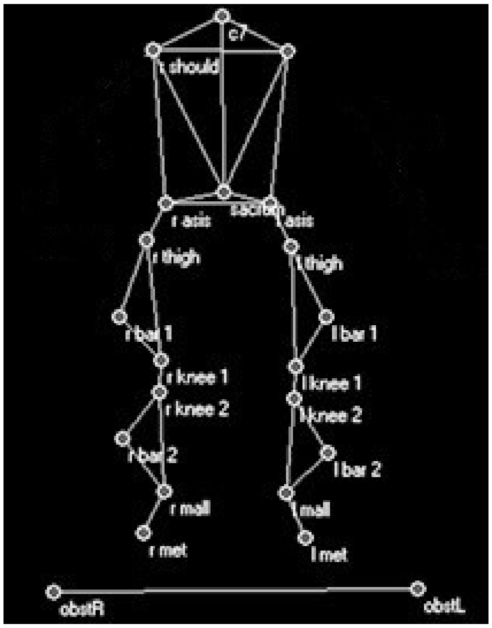
Table 2 - Overview of the subjects' characteristics (mean±SD); age, height, weight, and number of subjects.

	DS	*N*
Age (years)	21.6±7	25.1±2.4
Height (m)	1.52±0.08	1.68±0.07
Weight (kg)	56±9.1	60±7.5
Number of subjects	21	18

3.2 Acquisition and Instrumentation

The subjects walked along a walkway of approximately ten meters length in two conditions: plane walking (wlk) and walking with an obstacle (10% of the subject's height, obs). They performed walking in each condition three times. The obstacle was a wooden stick, which was supported by two supports placed laterally to the walkway. Ground reaction forces were sampled from two force platforms mounted in the middle of walkway. The tasks were acquired using quantitative movement analysis, composed of an optoelectronic system (Elite2002, BTS) with eight infrared cameras. The optoelectronic system records the three-dimensional coordinates of the markers through time. Markers were placed on the body according to Davis' protocol (Davis et al, 1991) and two markers were put respectively at the two ends of the obstacle to define the obstacle position relative to the subject during the movement (Figure 2)

Figure 2 - Davis' protocol for markers placement (Davis et al, 1991)



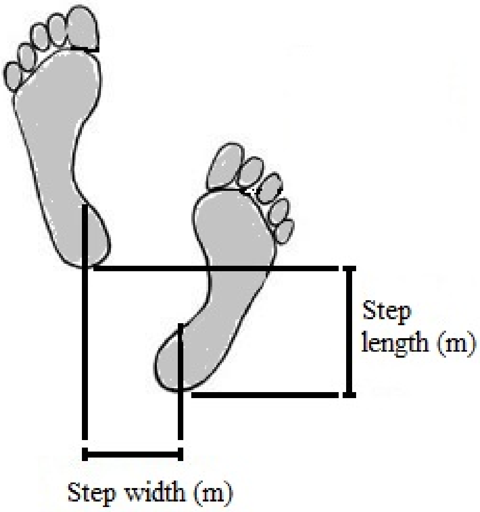
3.3 Parameters

In this study, some of the spatiotemporal parameters, mechanical energy parameters, and lower limb joints' work and power were computed. The spatiotemporal and mechanical energy parameters were computed from the markers' coordinates, while for computation of lower limb joints' power and work, the data from the force plates were collected too.

3.3.1 Spatiotemporal Parameters

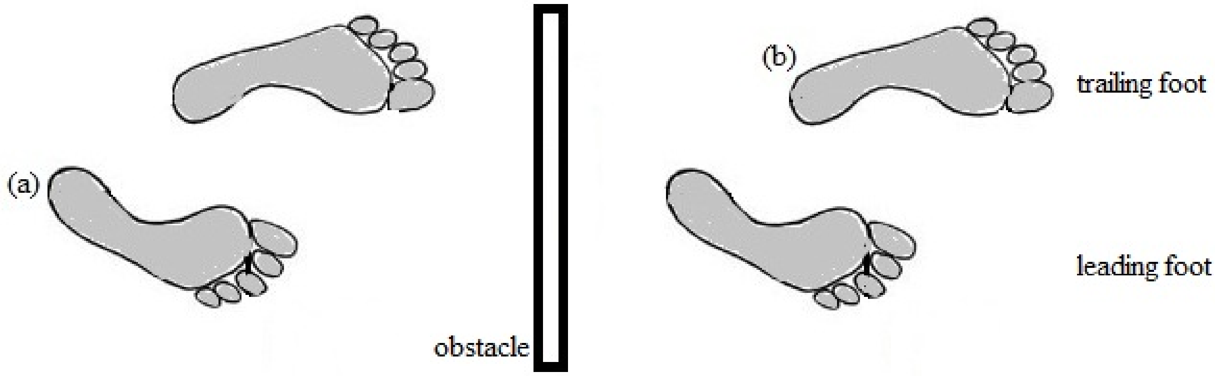
In this study, step length, step width, and mean velocity were computed from the markers' coordinates during walking. Step length and step width were respectively defined as the anteroposterior (AP) distance between two consecutive heel contacts of the feet and the mediolateral (ML) distance between two heel centers of two consecutive foot contacts (Figure 3).

Figure 3 - step length (AP distance between two consecutive heel contacts of the feet) and step width (ML distance between two heel centers of two consecutive foot contacts)



Mean velocity, which is an indicator of conservatism of the movement (chen et al, 1991), was defined as the average velocity of the marker on the sacrum during walking. Also, in obs condition, mean velocity was defined as the average velocity of the marker on the sacrum from the last heel strike of leading foot before the obstacle to the first heel strike of trailing foot after the obstacle (Figure 4).

Figure 4 - In obs, mean velocity was defined as the average velocity of the marker on the sacrum from the last heel strike of leading foot before the obstacle (a) to the first heel strike of trailing foot after the obstacle (b).



3.3.2 Mechanical Energy Parameters

An important component of gait analysis is energy computation, which gives a measure of the amount of energy required to walk over a given distance. Mechanical energy is introduced as the energy state of any limb segments or total body system at an instant in time. Mechanical energy consists of two main components, kinetic energy and potential energy (gravitational potential energy and elastic potential energy). In literature, different approaches such as CM approach, SSE approach, and integrated joint power approach, have been introduced to estimate mechanical energy (Van de Walle et al, 2011). In this study, CM approach has been used to estimate mechanical energy exchanges. In this regard, the following parameters were computed:

- maximum and minimum kinetic energy values (two max and two min in each gait cycle, Figure 5)
- maximum and minimum potential energy values (two max and two min in each gait cycle, Figure 6)
- total mechanical energy exchange
- the mechanical energy recovery

All of the above mentioned energy parameters have been normalized by the body weight (kg) and body height (m).

Figure 5 - Kinetic energy parameters (MinKE1, MaxKE1, MinKE2, MaxKE2)

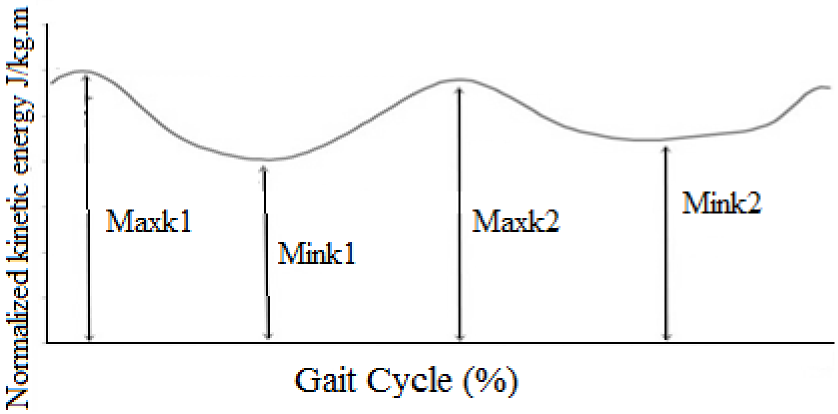
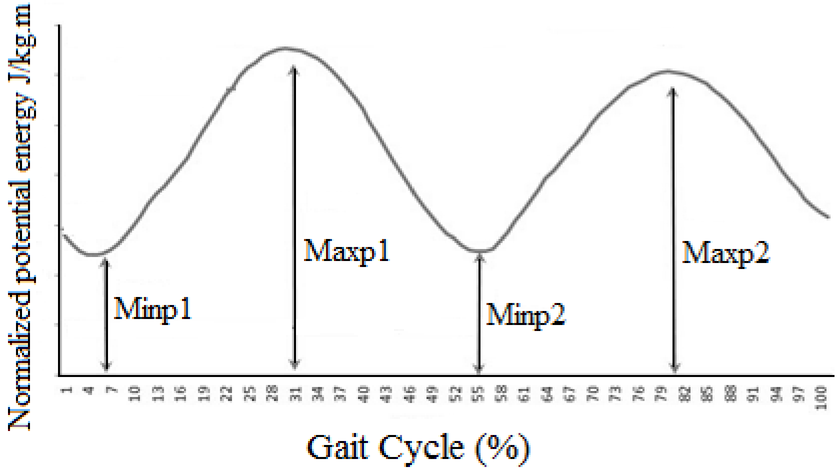


Figure 6 - Potential energy parameters (MinPE1, MaxPE1, MinPE2, MaxPE2)



The definitions of parameters, which are shown in the figures mentioned above, have been listed in Table 3. The differences between peaks have been presented in absolute values.

Table 3 - Potential and kinetic energy parameters' definition

	Parameter	*Description*
1	MaxPE1	Maximum Potential energy value in first peak
2	MaxPE2	Maximum Potential energy value in second peak
3	MinPE1	Minimum Potential energy value in first peak
4	MinPE2	Minimum Potential energy value in second peak
5	MaxKE1	Maximum Kinetic energy value in first peak
6	MaxKE2	Maximum Kinetic energy value in second peak
7	MinKE1	Minimum Kinetic energy value in first peak
8	MinKE2	Minimum Kinetic energy value in second peak
9	RangePE1	Difference between first maximum and minimum peak in Potential energy
10	RangePE2	Difference between second maximum and minimum peak in Potential energy
11	RangeKE1	Difference between first maximum and minimum peak in Kinetic energy
12	RangeKE2	Difference between second maximum and minimum peak in Kinetic energy

3.3.2.1 Kinetic Energy (KE)

For an object, the kinetic energy consists of two sentences; linear kinetic energy and rotational kinetic energy. We calculated KE as:

$$KE = {}^{1}/_{2}\, mv^{2} + {}^{1}/_{2}\, I\omega^{2}$$

where:

m is the mass of the object (translational inertia).

v is the velocity of the object.

I is the moment of inertia of the object about an axis passing through the CM, and perpendicular to the plane of motion (rotational inertia).

ω is the angular velocity.

In human walking, the second sentence is negligible compared to the first one (Winter et al, 1976). Since the rotational kinetic component can be ignored, the equation above can be expressed as follow:

$$KE = {}^{1}\!/_{2}\, mv^{2}$$

In this equation, v is the velocity of the CM in the AP direction, which is calculated from the position of the marker on the sacrum during walking.

In this study, the following kinetic energy parameters were calculated: MaxKE1, MaxKE2, MinKE1, MinKE2, RangeKE1, and RangeKE2. The definitions of these parameters were explained in Figure 5 and Table 3.

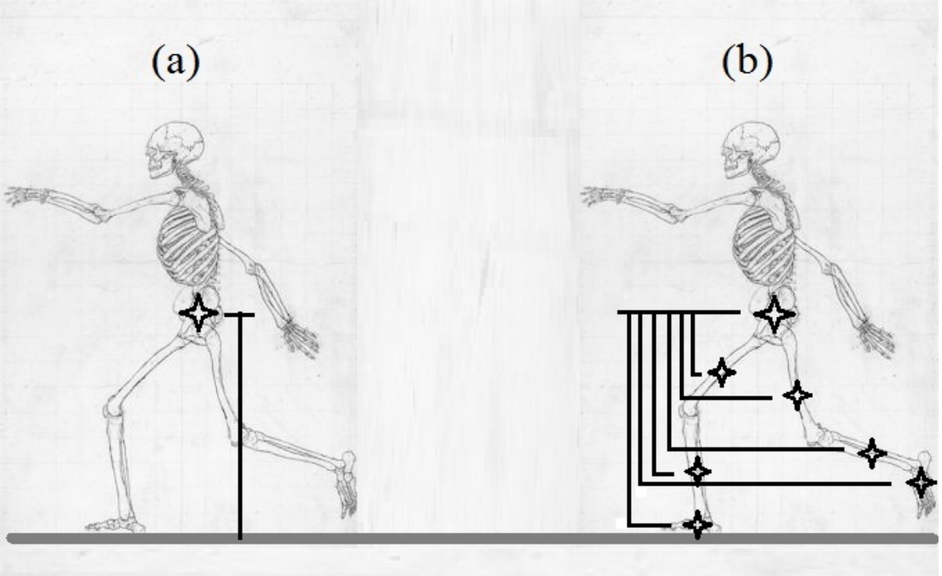
3.3.2.2 Potential Energy (PE)

Gravitational potential energy of an object depends on its mass, and its height or vertical movement. Thus, the gravitational potential energy is the energy stored in an object as the result of its vertical position or movement. Potential energy was calculated by the following equation:

$$PE = mg\Delta h$$

where m, g, and Δh are respectively represented the mass of the object, gravitational acceleration, and vertical displacement. Potential energy, in its turn, consists of two components. The first component refers to the change of potential energy due to vertical movements of the center of mass of body (CM). The second component takes into account the effect of vertical movements of the center of mass of lower limb segments (CM_{limbs}) respect to the CM (Figure 7).

Figure 7 - two components of potential energy (a) potential energy change due to vertical movement of the CM respect to ground (b) potential energy changes due to vertical movements of the center of mass of the lower limb (CM_{limbs}) segments respect to the CM.



In literature, three different methods have been introduced for calculation of vertical movement of the CM: the sacral marker method, the segmental analysis method, and the force platform method. The sacral marker method involved estimating vertical CM motion by tracking the position of a reflective marker that was placed on the sacrum of subjects as they walked. The body segmental analysis technique determined the vertical motion of the CM from a weighted average of the vertical positions of the centers of mass of individual body segments for each frame of kinematic data acquired during the data trial. The third technique involved calculating CM vertical motion through double integration of force platform data (Gard et al, 2004). In this study, the first method was used for calculation of vertical movement of the CM.

In this study, the following potential energy parameters were calculated: MaxPE1, MaxPE2, MinPE1, MinPE2, RangePE1, and RangePE2. The definitions of these parameters were explained in Figure 6 and Table 3.

Total mechanical energy, TME, equals the sum of potential energy and kinetic energy. So, it was calculated from the following equation:

$$TME = PE + KE = mg\Delta h + {}^{1}\!/_{2}\, mv^2$$

3.3.2.3 Mechanical Energy Recovery

As it was explained in the second chapter, in each gait cycle, some of the kinetic energy is converted into the gravitational potential energy, and consequently, some of the gravitational potential energy is converted back into the kinetic energy. Mechanical energy recovery shows how much of the mechanical energy (sum of potential energy and kinetic energy) can be recovered due to conversion of potential energy to kinetic energy and vice versa. Mechanical energy recovery was calculated from the following equation (Cavagna et al, 1976):

$$ER = \frac{(E_P + E_K) - E_{tot}}{E_P + E_K}$$

where:

$$E_P = nRangePE1 + nRangePE2$$

$$E_K = nRangeKE1 + nRangeKE2$$

$$E_{tot} = nRangeTME1 + nRangeTME2$$

And;

nRangePE1 is the normalized value of the difference between first maximum and minimum peak in potential energy in the gait cycle;

nRangePE2 is the normalized value of the difference between second maximum and minimum peak in potential energy in the gait cycle;

nRangeKE1 is the normalized value of the difference between first maximum and minimum peak in kinetic energy in the gait cycle;

nRangeKE2 is the normalized value of the difference between second maximum and minimum peak in kinetic energy in the gait cycle;

nRangeTME1 is the normalized value of the difference between first maximum and minimum peak in total mechanical energy in the gait cycle;

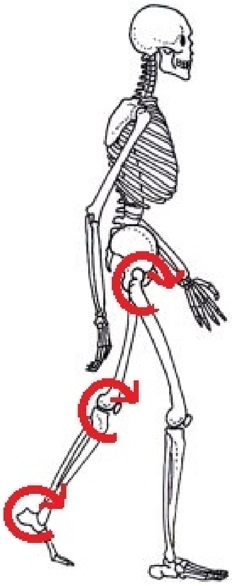
nRangeTME2 is the normalized value of the difference between second maximum and minimum peak in total mechanical energy in the gait cycle.

3.3.3 Power and Work

Net joint powers (P) were calculated as the product of net moments across the joint (M) and the relative angular velocity between the adjacent limb segments (ω). A positive power implied energy generation in muscles by contracting concentrically, and in contrast, a negative power signified the muscles absorbed energy by contracting eccentrically (Winter, 1983). Mechanical power is the indicator of the rate of change of mechanical energy.

$$P = M.\omega$$

Figure 8 - Moments across the lower limb joints (hip, knee, and ankle). Mechanical power depends on momentum and angular velocity



Also, the integral of power over time defines the work done. In other words, the area under the power curve provided the mechanical work generated or absorbed at the hip, knee, and ankle (Winter, 1983).

$$W = \int Pdt$$

Power parameters are normalized by the body weight (kg), when mechanical work parameters are normalized by the body height (m) and body weight (kg).

In this study, the following parameters were calculated: maximum and minimum values of generated and absorbed power at the ankle, maximum and minimum values of generated and absorbed power at the knee during the stance phase, and the range (the difference between maximum and minimum) of the mechanical work that was produced at the hip, knee, and ankle joints during plane walking.

The following table summarizes the main information about mechanical energy parameters, power, and work.

Table 4 - Mechanical energy, power, and work

Parameter	***Formula***	
Potential energy (PE)	$mg\Delta h$	Δh: vertical displacement of the CM
Kinetic energy (KE)	$mv^2/2$	v: mean velocity of the CM
Total mechanical energy (E_{tot})	$PE + KE$	
Mechanical energy Recovery (ER)	$(E_P + E_K) - E_{tot}/(E_P + E_K)$	
Power (P)	$M.\omega$	M, ω: Moment and angular velocity
Work (W)	$\int Pdt$	

3.4 Statistical Analysis

At the first, Kolmogorov-Smirnov test was used for evaluation of normality of the distribution of data. Then, a 2 conditions x 2 groups ANOVA was used to analyze the presence of statistically significant differences ($p\text{-value} < 0.05$) between the two groups (N and DS) in the two conditions (wlk and obs).

CHAPTER FOUR • RESULTS

All the subjects successfully completed the tasks. This chapter presents the obtained results. A sample of reports obtained is shown in APPENDIX A (for a healthy subject). In this chapter, the results are divided into three different subchapters; “spatiotemporal gait characteristics”, “mechanical energy”, and “power and work”.

4.1 Spatiotemporal Gait Characteristics

Table 5 shows the results of the spatiotemporal parameters (median, 25th and 75th percentile values) for plane walking and stepping over obstacle for both of N and DS groups.

Table 5 - Median (25th percentile, 75th percentile) values of the spatiotemporal parameters, wlk: plane walking condition, obs: stepping over obstacle

	wlk		*obs*		
	N	**DS**	**N**	**DS**	
Step Length (m)	0.586 (0.569,0.611)	0.422 (0.379,0.457)	0.603 (0.561,0.640)	0.415 (0.340,0.485)	§ *
Step Width (m)	0.152 (0.147,0.177)	0.199 (0.179,0.217)	0.210 (0.197,0.226)	0.242 (0.226,0.275)	§ * # +
Mean Velocity (m/s)	1.204 (1.125,1.265)	0.658 (0.598,0.730)	1.054 (1.003,1.166)	0.597 (0.470,0.672)	§ * #

§ significant difference between N-wlk and DS-wlk (p-value < 0.05)
* significant difference between N-obs and DS-obs (p-value < 0.05)
# significant difference between N-wlk and N-obs (p-value < 0.05)
+ significant difference between DS-wlk and DS-obs (p-value < 0.05)

For DS, compared to N, reduced and more variable step length was found in both conditions. Moreover, percentile ranges (the difference between 25th and 75th percentiles) of step length was higher for DS compared to N in both conditions, which means DS walked and stepped over obstacle with more variable step length compared to N. DS completed their tasks in wlk and obs conditions with respectively 28% and 31% decrease in step length respect to N subjects. There is no statistically significant difference in step length values between wlk and obs within groups for both of DS and N subjects.

For both of wlk and obs conditions, results showed that DS walked with greater step width compared to N. In addition, DS walked with increased step width variability. Statistical comparisons revealed that, for DS compared to N, step width was greater with 30% and 15% in wlk and obs conditions, respectively. Also, for both groups of subjects, step width was greater in obs than wlk (for DS with 22%, for N with 38%)

As shown in Table 5 above, DS walked with lower mean velocity compared to N in both conditions. For wlk and obs conditions, velocity was decreased respectively by 45% and 43% from N to DS. Also, while there is no statistically significant difference in DS's velocity between the two conditions, N walked slower by 12% in obs condition than wlk condition.

4.2 Mechanical Energy Parameters

In Table 6, the results of the mechanical energy parameters (median, 25th and 75th percentile values) are listed for plane walking and stepping over obstacle. All of these parameters are normalized (start with "n" in the table below) by the body height (m) and body weight (kg).

Table 6 - Median (25th percentile, 75th percentile) values of the mechanical energy parameters, wlk: plane walking condition, obs: stepping over obstacle

	wlk		***obs***		
	N	**DS**	**N**	**DS**	
nMaxPE1 (J/kg.m)	5.467 (5.418,5.541)	5.491 (5.403,5.608)	5.475 (5.397,5.609)	5.394 (5.273,5.597)	
nMaxPE2 (J/kg.m)	5.466 (5.418,5.540)	5.516 (5.411,5.612)	5.517 (5.492,5.656)	5.552 (5.371,5.666)	
nMinPE1 (J/kg.m)	5.255 (5.175,5.331)	5.355 (5.189,5.453)	5.265 (5.202,5.349)	5.257 (5.109,5.490)	
nMinPE2 (J/kg.m)	5.256 (5.135,5.312)	5.330 (5.190,5.422)	5.257 (5.173,5.320)	5.258 (5.002,5.362)	
nRangePE1 (J/kg.m)	0.214 (0.194,0.244)	0.173 (0.114,0.198)	0.224 (0.185,0.264)	0.187 (0.143,0.232)	§ *
nRangePE2 (J/kg.m)	0.239 (0.200,0.256)	0.197 (0.159,0.228)	0.318 (0.272,0.357)	0.309 (0.234,0.381)	* # +
nMaxKE1 (J/kg.m)	0.647 (0.567,0.720)	0.303 (0.249,0.380)	0.565 (0.457,0.685)	0.252 (0.177,0.331)	§ *
nMaxKE2 (J/kg.m)	0.662 (0.582,0.720)	0.305 (0.253,0.395)	0.567 (0.456,0.794)	0.252 (0.177,0.346)	§ *
nMinKE1 (J/kg.m)	0.372 (0.315,0.413)	0.102 (0.090,0.167)	0.230 (0.116,0.274)	0.076 (0.012,0.100)	§ * # +
nMinKE2 (J/kg.m)	0.332 (0.294,0.398)	0.093 (0.075,0.150)	0.141 (0.099,0.206)	0.019 (0.005,0.041)	§ * # +
nRangeKE1 (J/kg.m)	0.272 (0.251,0.303)	0.195 (0.139,0.210)	0.324 (0.226,0.410)	0.168 (0.110,0.241)	§ *
nRangeKE2 (J/kg.m)	0.303 (0.278,0.341)	0.210 (0.164,0.238)	0.437 (0.308,0.559)	0.221 (0.172,0.312)	§ * #
ER	78% (68% , 85%)	78% (72% , 84%)	74% (67% , 83%)	72% (60% , 84%)	

§ significant difference between N-wlk and DS-wlk (p-value < 0.05)
* significant difference between N-obs and DS-obs (p-value < 0.05)
# significant difference between N-wlk and N-obs (p-value < 0.05)
+ significant difference between DS-wlk and DS-obs (p-value < 0.05)

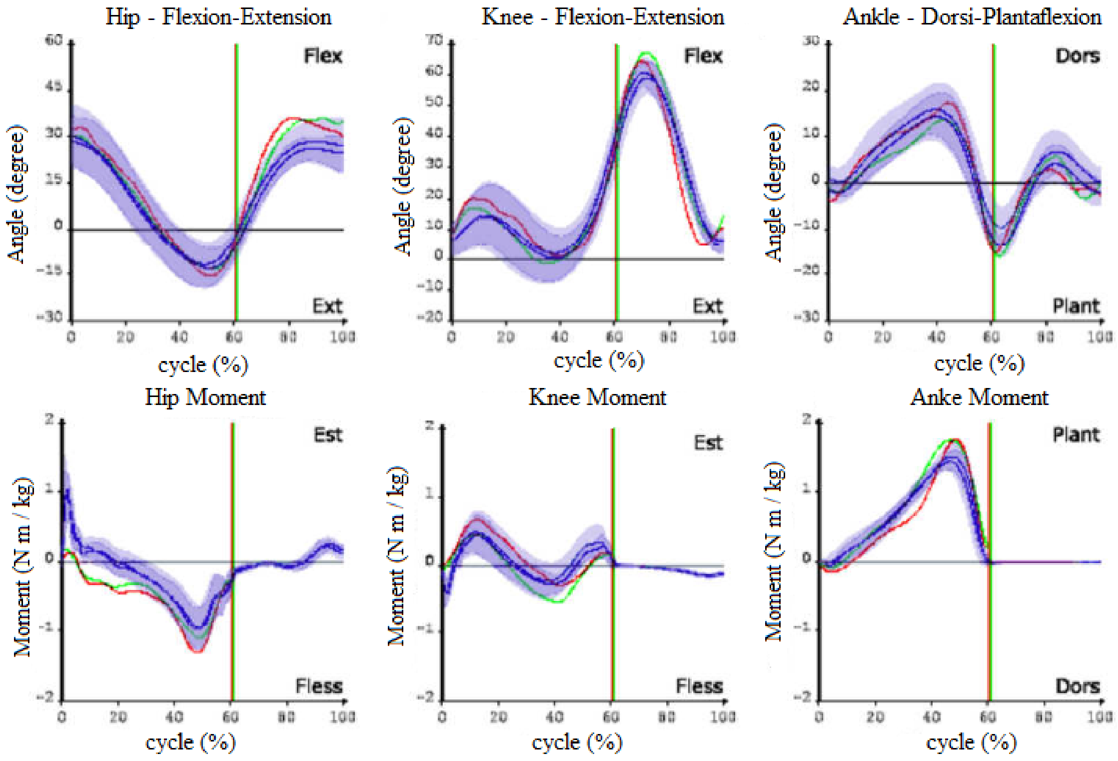
As shown in the table, except nRangePE1 and nRangePE2, there are no significant differences in the other potential energy parameters between DS and N within conditions. nRangePE1 and nRangePE2 were higher for N respect to DS in obs condition. In addition, nRangePE1 was higher for N respect to DS in wlk condition too. On contrary, all of the kinetic energy parameters between DS and N differ in both of wlk and obs conditions. In other words, DS walked with lower kinetic energy than N in both conditions. Also, nMinKE1 and nMinKE2 parameters are different between two conditions for both of two groups.

In the next step, the potential energy and kinetic energy parameters have been used to calculate the mechanical energy recovery (ER), which has a clear clinical meaning for comparing the groups and conditions. ER parameter reveals that which group of subjects can better recover the mechanical energy that was consumed in the gait cycle. In both conditions, DS walked with the same ER value respect to N. Also, ER value did not change across conditions for both groups (N and DS).

4.3 Power and Work

In this study, in addition to the above mentioned parameters, the powers generated or absorbed at the knee, hip, and ankle were calculated and analyzed. Mechanical work during stance was then quantified by integrating the power-time curves. This analysis was just performed for wlk condition, because some of the subjects did not put correctly their feet on the force platforms in obs condition and their force plates' data were not usable. So, the comparison between to conditions within groups was not applicable for power and work parameters. For this purpose, flex-extension and moment curves at the hip, knee, and ankle joints were obtained. As an example, these curves are shown in Figure 9, for one of the healthy subjects.

Figure 9 - Sagittal kinetic of hip, knee and ankle for a healthy subject (red line: left foot, green line: right foot)



In Table 7, the absolute values of mechanical power and work parameters (median, 25th and 75th percentile values) are presented for plane walking. Power parameters are normalized (start with "n" in the following table) by the body weight (kg), when mechanical work parameters are normalized by the body height (m) and body weight (kg).

Table 7 - Absolute values of median (25th percentile, 75th percentile) of the power and work parameters for right foot.

	N	*DS*	
nMax generated power at ankle (W/kg)	0.0845 (0.0628,0.0923)	0.0271 (0.0202,0.0401)	§
nMin absorbed power at ankle (W/kg)	0.0124 (0.0088,0.0180)	0.0103 (0.0077,0.0141)	

	N	*DS*	
nMax generated power stance at knee (W/kg)	0.7143 (0.5382,1.0844)	0.5093 (0.2612,0.7191)	§
nMin absorbed power stance at knee (W/kg)	0.9574 (0.6713,1.2171)	1.0063 (0.7515,1.3489)	
nRange ankle work (kJ/kg.m)	0.0147 (-0.0033,0.0233)	0.0280 (0.0222,0.0381)	§
nRange knee work (kJ/kg.m)	0.0104 (0.0014,0.0279)	0.0002 (-0.0153,0.0049)	§
nRange hip work (kJ/kg.m)	0.0659 (0.0397,0.1018)	0.1094 (0.0931,0.1385)	§

§ significant difference between N-wlk and DS-wlk (p-value < 0.05)

In wlk condition, the powers generated at the ankle and knee joints were higher in N respect to DS. Also, compared to N group, subjects with DS produced greater work at the ankle and hip. At the knee, while N subjects produced levels of work similar to those at the ankle, the produced work equals almost zero in subjects with DS.

CHAPTER FIVE • CONCLUSION

In the first part of the chapter, the results are discussed in three different paragraphs: “spatiotemporal parameters”, “mechanical energy recovery”, and “power and work”. Then, in the next part, all results are concluded; and at last, the limitations of the present study and some suggestions for the future researches are discussed.

5.1 Discussion

5.1.1 Spatiotemporal Parameters

In all conditions, mean velocity was significantly lower in DS respect to N (DS's mean velocity was about half of N's in both conditions). This result is in agreement with previous studies (Rigoldi et al, 2010; Vimercati et al, 2011). As Rigoldi et al (2010) have concluded it is related to perceived instability in DS. N subjects walked slower in the presence of an obstacle respect to plane walking, whereas subjects with DS, who already walked slower, completed the tasks with the same mean velocity in both conditions. In addition, both groups showed higher variability in velocity (demonstrated by higher percentile ranges) in obs condition than wlk condition. These results suggest that, however N decreased their mean velocity to analyze the new situation and adjust themselves with it, DS walked with similar velocities across conditions. When N reduced their mean velocity in order to have a safe movement over the obstacle, DS did not reduce their velocity in obs respect to wlk. This can be one of the probable reasons of greater falling risk in DS compared to N.

DS walked with greater step width in both conditions (wlk and obs). Moreover, their step width values were more variable (demonstrated by higher percentile ranges) in comparison with those for N. Altogether, both groups of subjects showed higher values of step width in obs conditions respect to wlk condition. It is possible to say that, DS walk with wider step width in order to increase their stability during plane walking and stepping over obstacle. Compared to N, step length values were reduced and more variable in subject with DS in all conditions. Our results confirmed what Vimercati et al (2011) had found through their study. On the other hand, both groups maintained their step length unvaried across conditions (wlk vs obs), with shorter step length in DS. Judge et al (1996) reported that step length has an inverse relationship with falling risk. The result of their study in conjunction with our results can provide another probable reason of higher falling risk in DS. In addition, a short step reduces the biomechanical demands (smaller AP momentum in joints due to smaller AP force during landing (King et al, 2005)), which may be preferential for subjects with DS who have reduced motor and physical capacities compared to N subjects.

5.1.2 Mechanical Energy Recovery

According to the theory of pendulum-like mechanism for walking, there are two maximums and minimums for kinetic energy in each gait cycle. In this study, these maximums and minimums were calculated (MinKE1, MinKE2, MaxKE1, MaxKE2). They were then normalized by the body weight (kg) and body height (m) (nMinKE1, nMinKE2, nMaxKE1, nMaxKE2). In both wlk and obs conditions, kinetic energy parameters were greater in N than DS. Subjects' kinetic energies were calculated from $KE = mv^2/2$. From this formula, kinetic energy depends on subjects' weight and their mean velocity. Therefore, normalized kinetic energy just depends on the mean velocity. Subjects with DS walked slower than N. So, their lower mean velocity resulted in smaller kinetic energy values compared with N.

On contrary, there is no statistically significant difference in max and min values of potential energy exchange between the two groups of subjects neither in wlk nor in obs. Potential energy exchange was calculated from $PE = mg\Delta h$. Then, it was normalized by the body height and body weight. Therefore, the normalized potential energy parameters just depends on the amount of Δh. Δh is the vertical displacement of the CM. So, from our results, it can be interpreted that the displacement of the CM (Δh) in both groups is the same in all conditions. These results demonstrated the similar CM excursion between two groups. However, max and min values of potential energy exchange were similar between N and DS, the normalized ranges of potential energy exchanges (nRangePE1 and nRangePE2) is greater for N compared to DS.

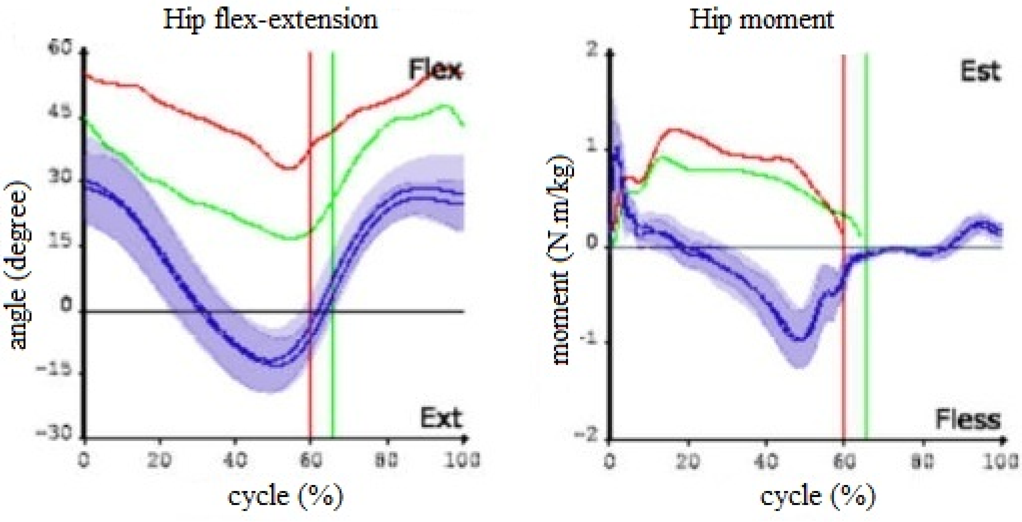
Mechanical energy recovery was similar for both of N and DS groups either during plane walking or stepping over obstacle. The formula that was used for computation of ER, consists of three main components: the range of potential energy exchange (nRangePE1 and nRangePE2), the range of kinetic energy (nRangeKE1 and nRangeKE2), and the range of total mechanical energy of the body (nRangeT1 and nRangeT2). In its turn, kinetic energy depends on mean velocity, and potential energy exchange depends on the vertical displacements of the CM. During both conditions, DS walked with lower velocity (resulting lower kinetic energy) and lower E_p (sum of nRangePE1 and nRangePE2) than N. On the other hand, for DS, the amount of E_{tot} (sum of nRangeT1 and nRangeT2) was smaller compared to N too. The interaction of three above mentioned sets of parameters results in very close ER values for both groups. In addition, each group had similar ER values across different conditions. It is possible to use the same line of reasoning to explain this event. As mentioned earlier, in each gait cycle, potential energy and kinetic energy are converted to each other like the mechanism of a hanging and inverted pendulum system. During each gait cycle, some of mechanical energy (sum of potential and kinetic energies) is converted into other forms of energy such as thermal energy (waste), and some can be recovered. ER is an energy parameter which indicates how much of total mechanical energy can be recovered during a gait cycle (Cavagna et al, 1976). Although DS used different strategies to walk respect to those used by healthy subjects (i.e. different velocity, step width, step length, and etc.), our results suggest that they could recover the same percentage of consumed mechanical energy with N during a gait cycle.

5.1.3 Power and Work

In wlk condition, power generated in the ankle was significantly higher in N than in DS. King et al (2005) suggested in their study that shortening the step length may have reduced AP momentum across the joints. It is confirmed by our study, as DS had shorter step length, and their joint momentums became smaller than in N subjects.

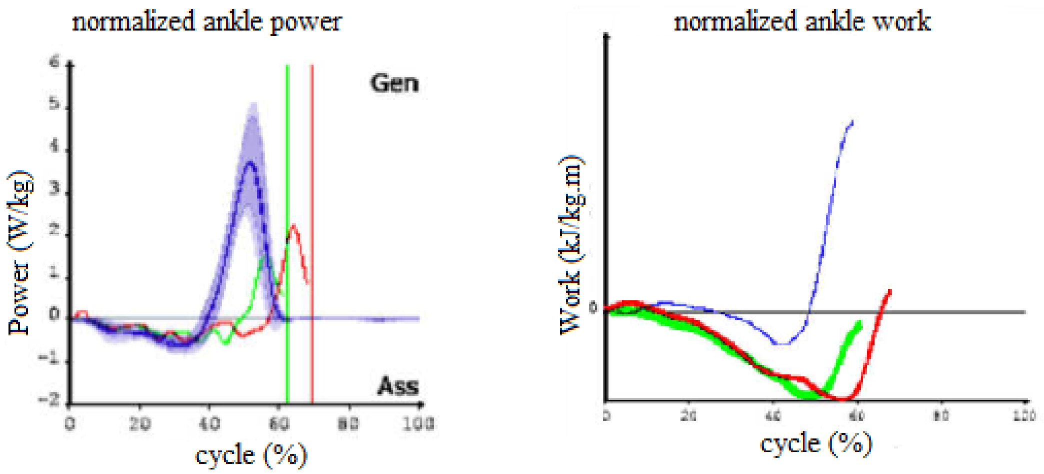
As mentioned earlier, power depends on the momentum across the joint and the relative angular velocity between the adjacent limb segments. According to our graphs, the angular velocity had more or less the same sign in both of N and DS, but the moments were opposite-sign for DS respect to N, result in the power with the opposite-sign for DS compared to N at the knee and hip joints. As an example, for one of the subjects with DS, the curves of hip flexion-extension and moment across the hip are plotted in Figure 10. In this figure, the green and red curves respectively belong to the right foot and left foot of the subject with DS, while the blue curve is for a healthy subject. The slope of flexion-extension curve equals the angular velocity. As seen in the left diagram, the slopes of flexion-extension curves have approximately the same sign for the subject with DS and the healthy person (from 0% to 50% of gait cycle: negative, from 50% to the end of gait cycle: positive). But as mentioned above and shown in the right diagram, the moments are opposite-sign for DS compared to N. This difference in joint momentums in subjects with DS leads to opposite-sign values of power generated in the joint. The same line of reasoning can be used to explain the behavior of power generated at the knee.

Figure 10 - Hip power depends on its angular velocity (slope of the flex-extension curve) and moment of hip; green line: DS right foot, red line: DS left foot, blue: Normal subject



Work is the integral of power over time. While DS produced greater amounts of work at the ankle and hip respect to N, the work produced at the knee was lower for them. In addition, for both groups, mechanical work produced at the knee was minimal compared to that at the ankle and hip joints, reinforcing the findings that the knee muscles mainly absorb, rather than generate energy (Winter, 1983). Normalized ankle power and work diagrams for one of the subjects with DS is shown in Figure 11. In these graphs, the red and green lines correspond respectively to the left foot and right foot of subject DS, where the blue one corresponds to N. As seen in figure, by increase or reduction of the area under the power curves, mechanical energy generated or absorbed at the ankle increases or reduces, respectively.

Figure 11 - Work is the integral of power over time; (a) normalized ankle power, and (b) normalized ankle work. green line: DS right foot, red line: DS left foot, blue: Normal subject



(a) (b)

5.2 Conclusion

Based on literature, reduced step length and velocity and increased step width are well-known features of DS walking pattern, which were confirmed by our study as well. In both wlk and obs conditions, DS showed higher value in step width and lower value and higher variability in velocity and step length. Though reduced velocity resulted in lower kinetic energy in DS, they showed similar potential energy exchanges with N during all conditions. In all conditions, DS subjects registered the same value of ER as N. It shows that DS could recover the same amount of consumed mechanical energy with N during a gait cycle. However flexion-Extension curves of the ankle and knee had similar behaviors, the ankle and knee moment diagrams revealed that, in DS respect to N, opposite-sign moments were applied at these joints. This is a characterizing feature of DS that results in different powers generated and works produced across lower limb's joints between DS and N groups.

5.3 Future Works and Limitations of the Study

However, in this study, almost numerous groups of subjects (both of DS and N) were considered, but the number of subjects (sample size) can be increased even more in order to obtain a better precision. In addition, subjects can be divided into different age group categories to decrease the variability within a group of subjects. Also, different age categories make it possible to track the motor ability development of the subjects with DS during lifespan.

As described in chapter three, subjects performed walking in each condition three times. Several repetitions can cause that subjects become familiar with the task and this modification can lead to minimum EE (Berard & Vallis, 2006). So, this walking strategy modification can reduce the precision of the study. On the other hand, with the same line of reasoning, it is possible to study the learning process in subjects with DS by increasing the number of task repetitions.

Speed is one of the factors that affect gait variables. In this study, the participants were asked to walk at a comfortable speed. That allowed us to determine how fast the participants walk and step over obstacle. In literature, it is suggested that walking should be tested under a range of speeds, rather than just self-selected speed (Beaman et al, 2010). It can be considered for future works.

Finally, in this study, mechanical energy exchanges were calculated in healthy subjects and subjects with Down syndrome. Since mechanical approaches don’t take into account all sources of EE, in the future studies, metabolic EE can be assessed and used for comparison between groups and different walking conditions for the same group. The results of metabolic EE assessment can be compared with our results in order to perform a more precise analysis.

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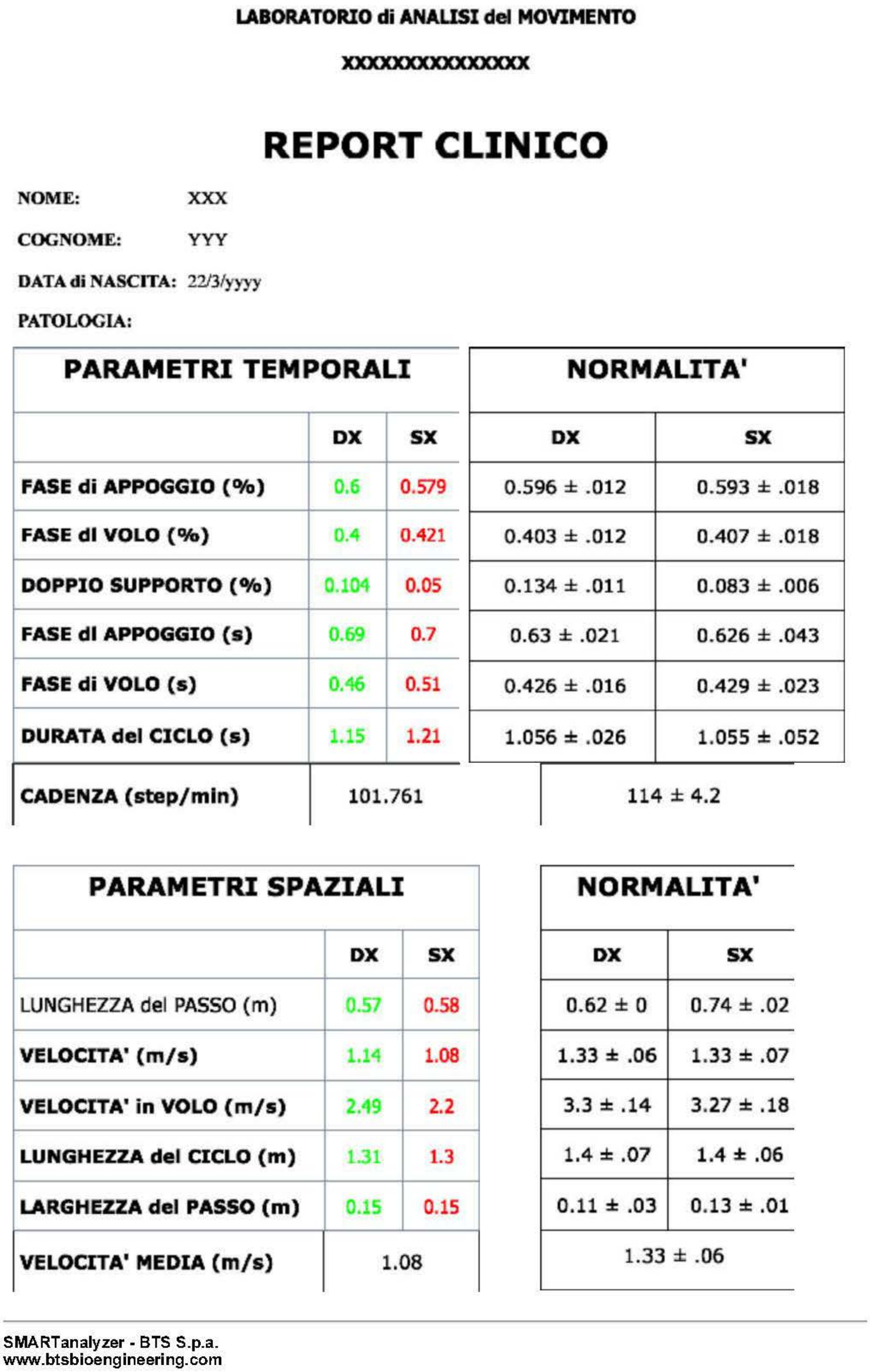
APPENDIX A



LABORATORIO di ANALISI del MOVIMENTO

XXXXXXXXXXXXXXX

REPORT CLINICO

NOME: XXX

COGNOME: YYY

DATA di NASCITA: 22/3/yyyy

PATOLOGIA:

PARAMETRI TEMPORALI			NORMALITA'	
	DX	SX	DX	SX
FASE di APPOGGIO (%)	0.6	0.579	0.596 ± .012	0.593 ± .018
FASE dI VOLO (%)	0.4	0.421	0.403 ± .012	0.407 ± .018
DOPPIO SUPPORTO (%)	0.104	0.05	0.134 ± .011	0.083 ± .006
FASE dI APPOGGIO (s)	0.69	0.7	0.63 ± .021	0.626 ± .043
FASE di VOLO (s)	0.46	0.51	0.426 ± .016	0.429 ± .023
DURATA del CICLO (s)	1.15	1.21	1.056 ± .026	1.055 ± .052
CADENZA (step/min)	101.761		114 ± 4.2	

PARAMETRI SPAZIALI			NORMALITA'	
	DX	SX	DX	SX
LUNGHEZZA del PASSO (m)	0.57	0.58	0.62 ± 0	0.74 ± .02
VELOCITA' (m/s)	1.14	1.08	1.33 ± .06	1.33 ± .07
VELOCITA' in VOLO (m/s)	2.49	2.2	3.3 ± .14	3.27 ± .18
LUNGHEZZA del CICLO (m)	1.31	1.3	1.4 ± .07	1.4 ± .06
LARGHEZZA del PASSO (m)	0.15	0.15	0.11 ± .03	0.13 ± .01
VELOCITA' MEDIA (m/s)	1.08		1.33 ± .06	

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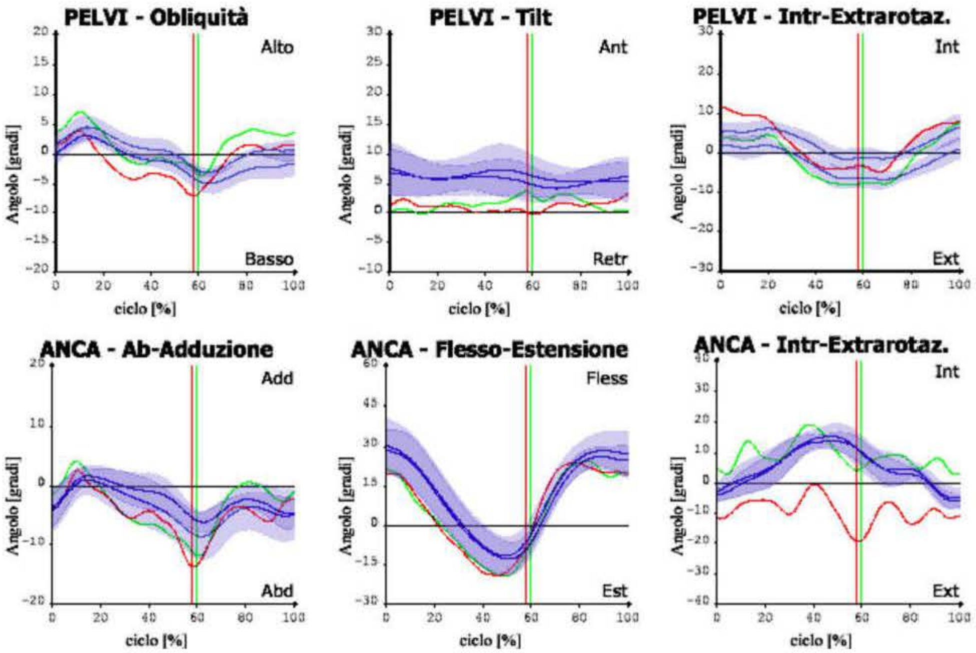
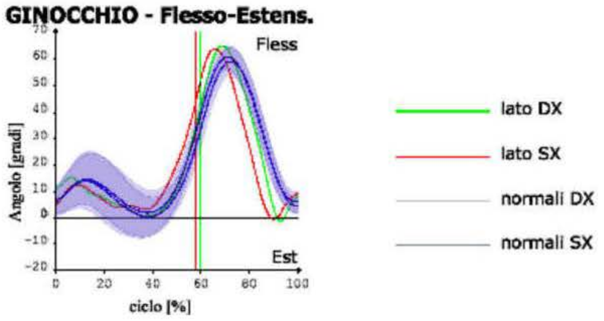
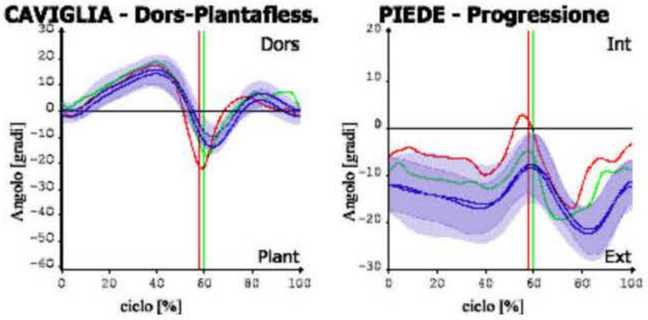
CINEMATICA MEDIA: DX su SX

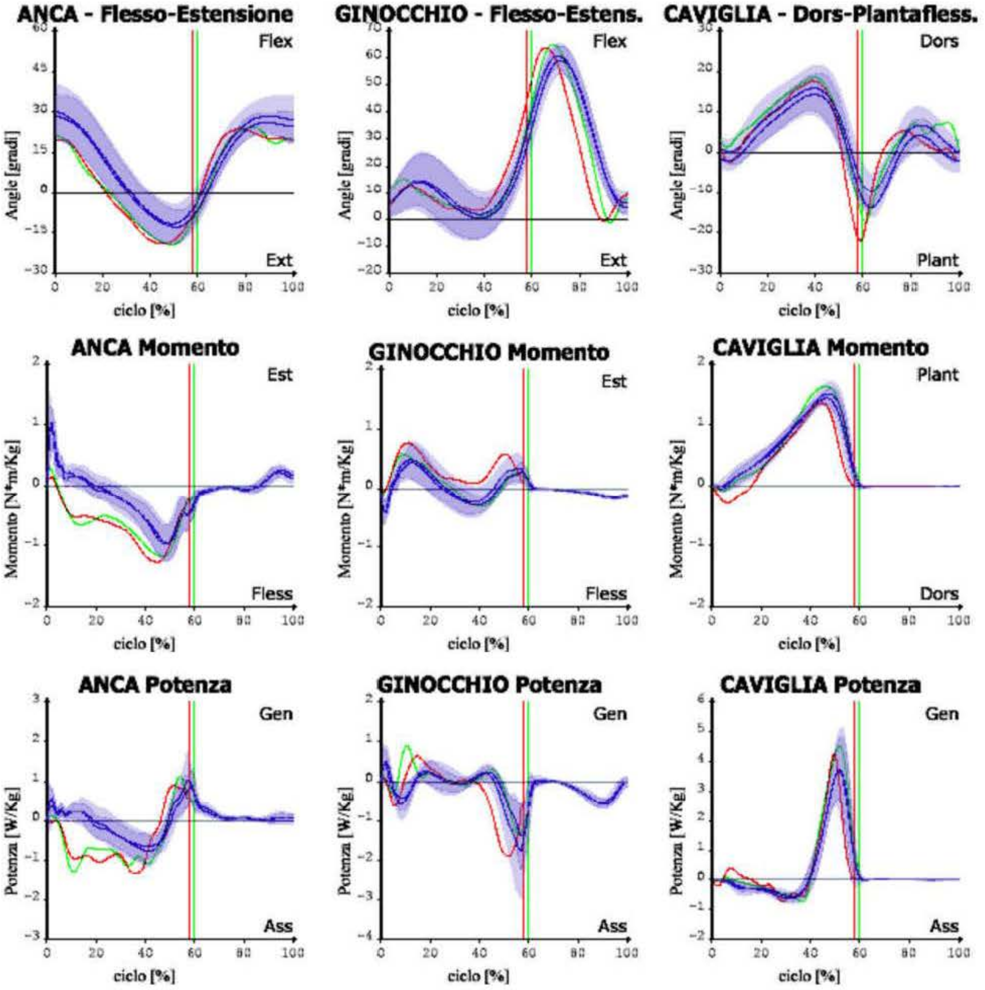




Angoli di Offset		
	DX	SX
Pelvi Obliquità	1.129	-1.129
Pelvi Rotazione	2.793	-2.793
Pelvi Tilt	8.449	8.449
Anca Abd-Add	-6.015	-7.506
Anca Intra-Extr	18.277	-3.901
Anca Fless-Est	-1.923	-3.762
Ginocchio Fless-est	-4.988	-7.697
Piede Progressione	-8.254	-12.351
Caviglia Dors-Plant	11.362	5.865



CINETICA SAGITTALE MEDIA: DX su SX



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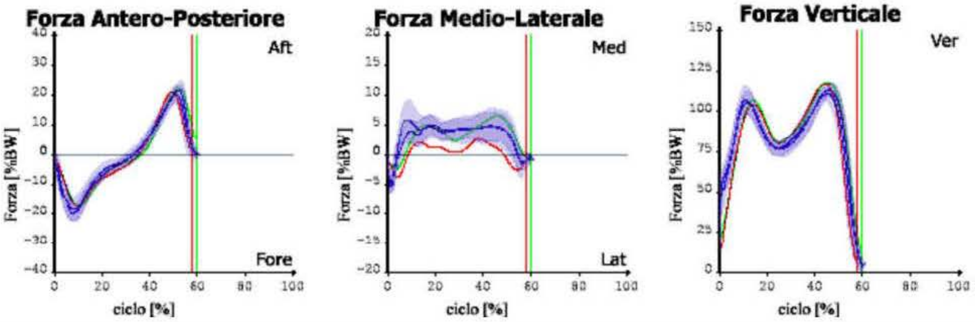

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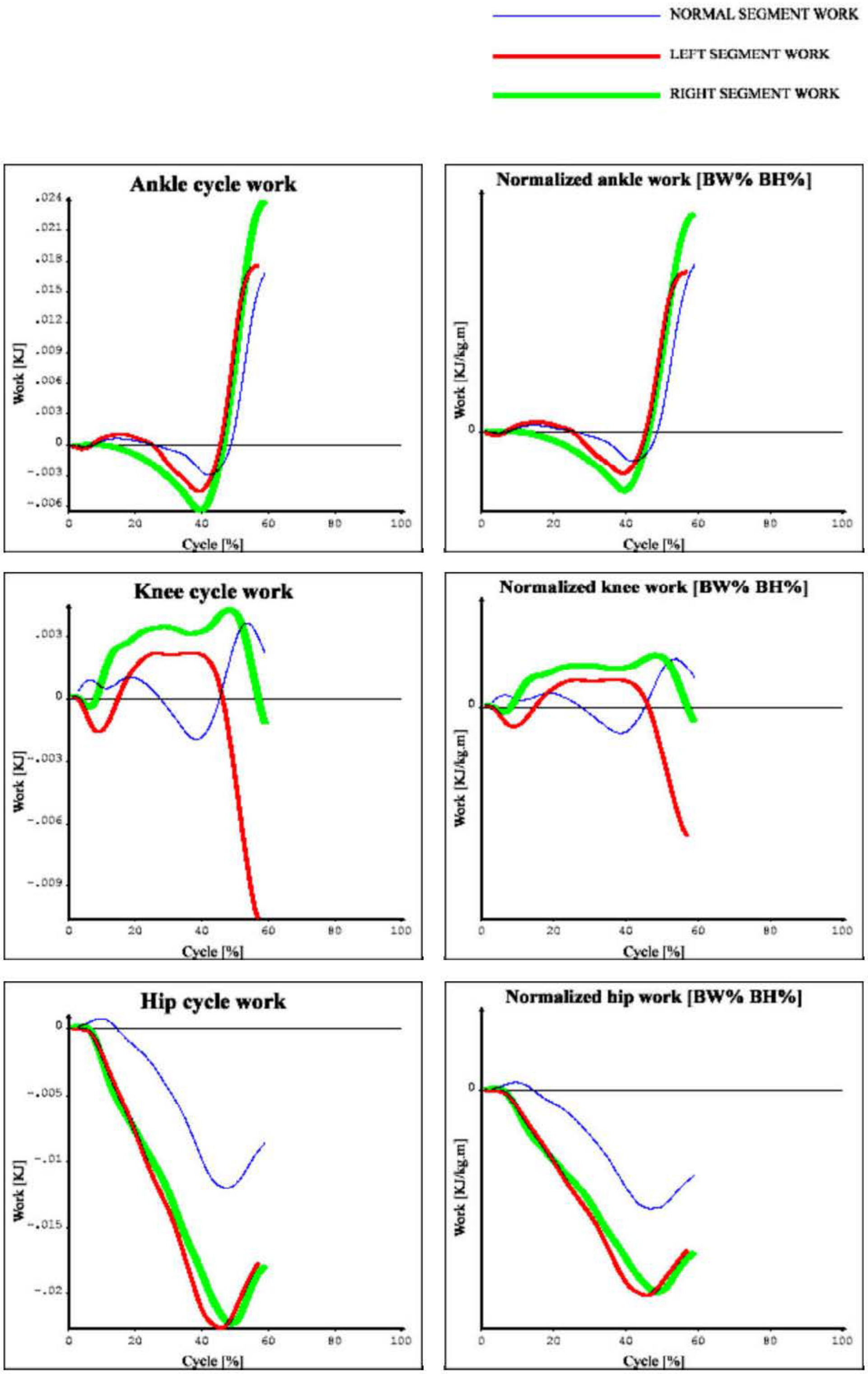
NORMAL SEGMENT WORK
LEFT SEGMENT WORK
RIGHT SEGMENT WORK
Ankle cycle work
Work [KJ]
.024
.021
.018
.015
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.009
.006
.003
0
-.003
-.006
0
20
40
60
80
100
Cycle [%]
Normalized ankle work [BW% BH%]
Work [KJ/kg.m]
0
Cycle [%]
Knee cycle work
Work [KJ]
.003
0
-.003
-.006
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Cycle [%]
Normalized knee work [BW% BH%]
Work [KJ/kg.m]
Cycle [%]
Hip cycle work
Work [KJ]
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Cycle [%]
Normalized hip work [BW% BH%]
Work [KJ/kg.m]
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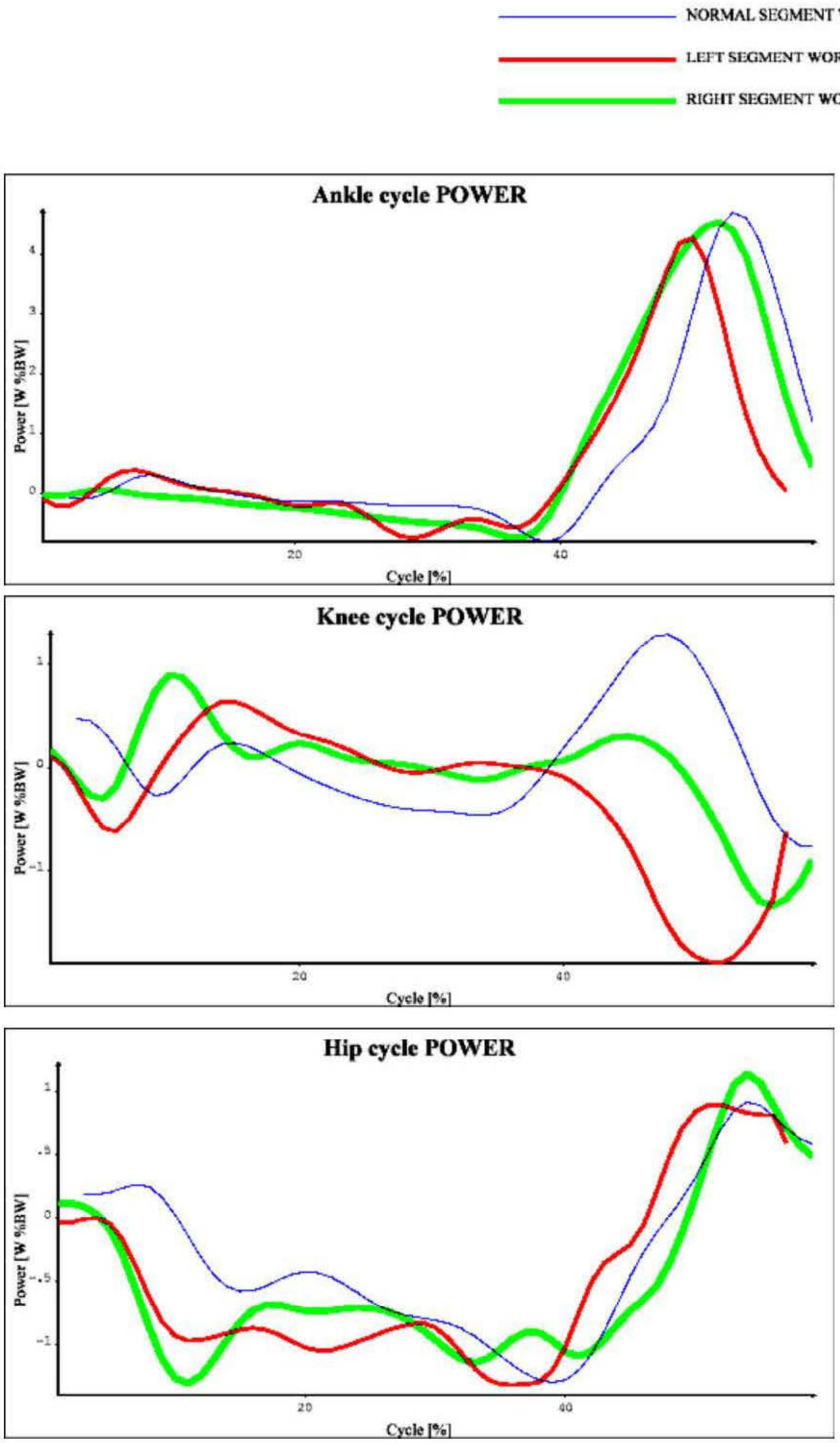
NORMAL SEGMENT WORK
LEFT SEGMENT WORK
RIGHT SEGMENT WORK
Ankle cycle POWER
Power [W %BW]
4
3
2
1
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20
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Cycle [%]
Knee cycle POWER
Power [W %BW]
1
0
-1
20
40
Cycle [%]
Hip cycle POWER
Power [W %BW]
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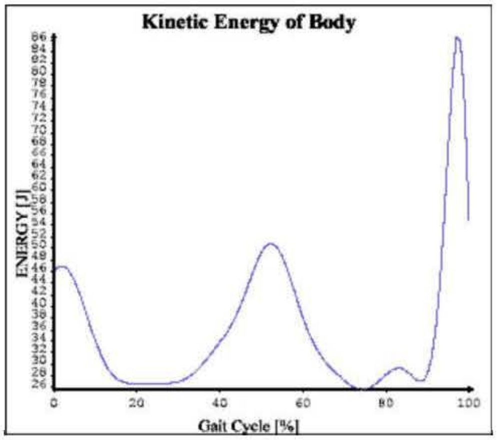
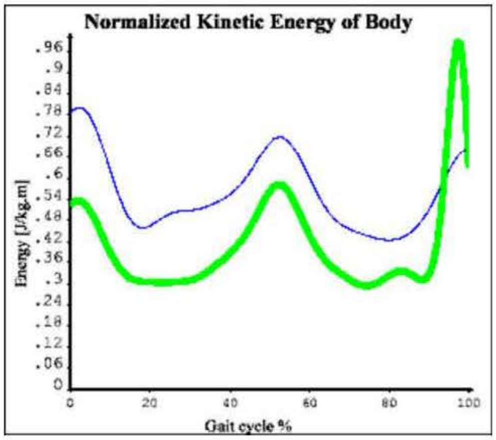
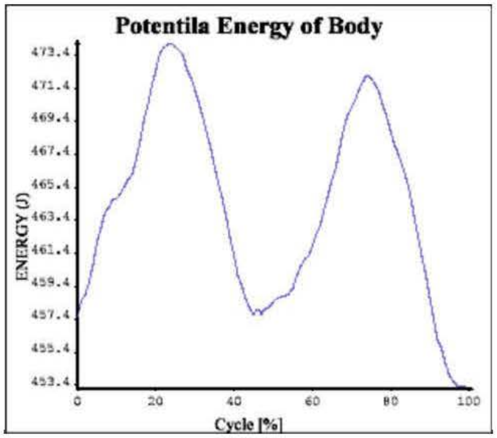
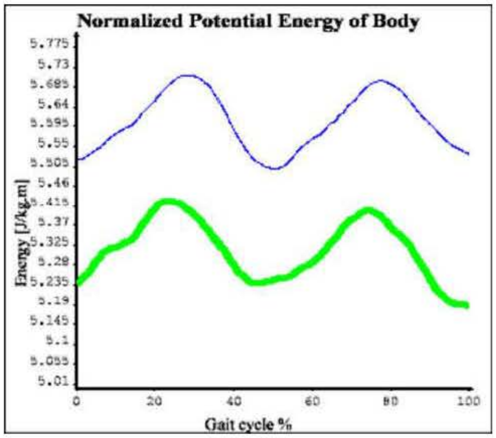
🟦 NORMAL ENERGY

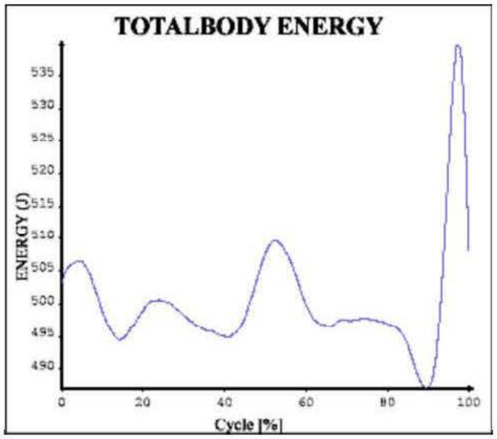
🟩 SUBJECT ENERGY

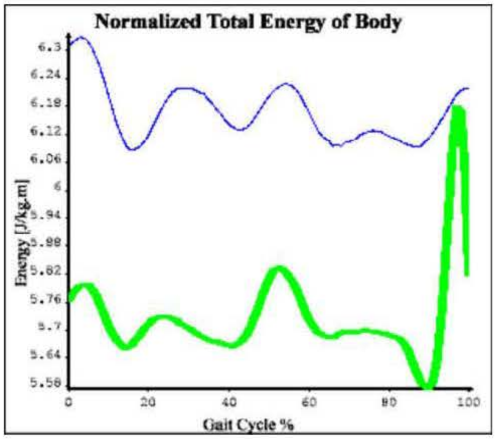


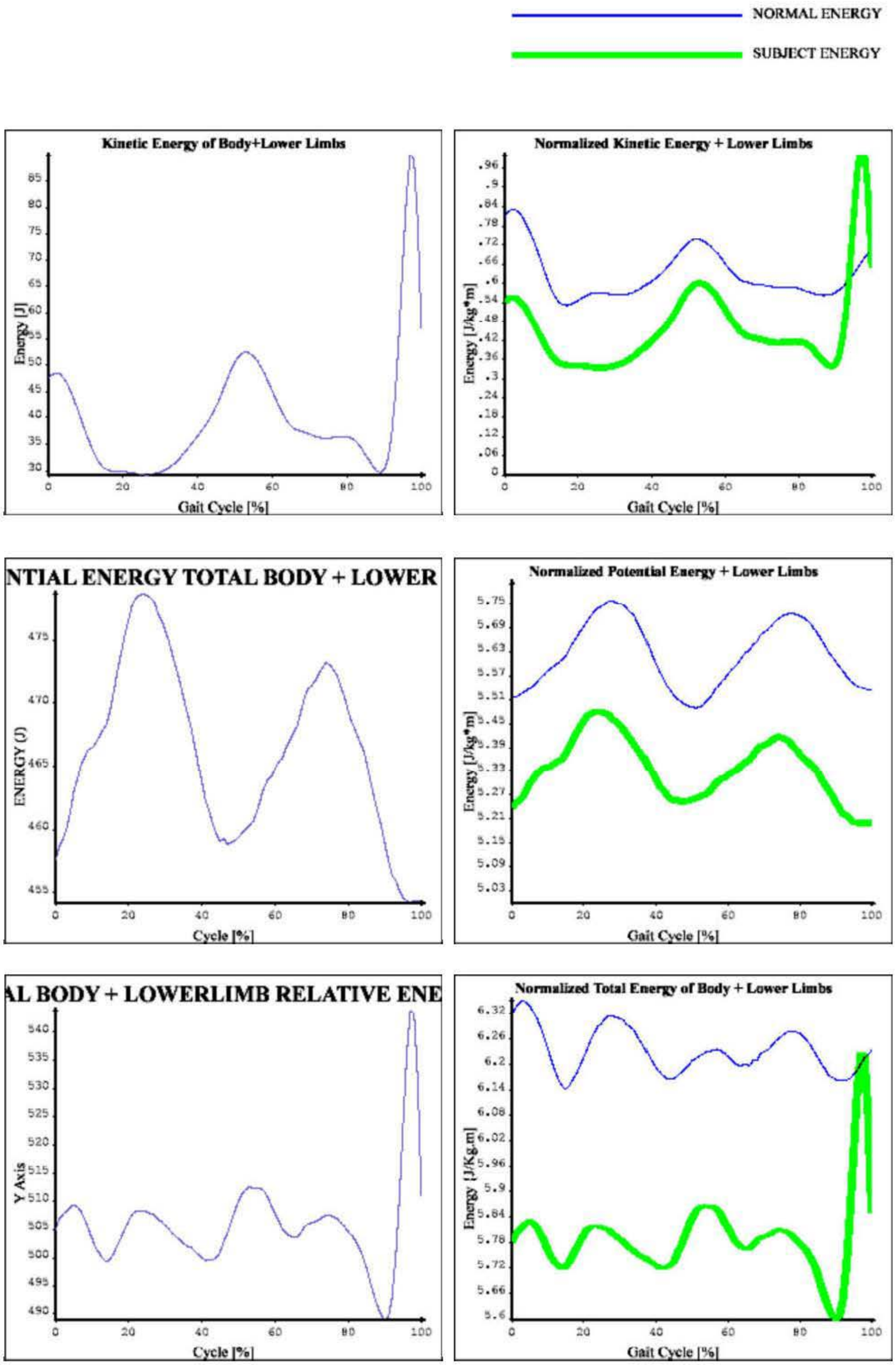












Kinetic Energy of Body+Lower Limbs

Normalized Kinetic Energy + Lower Limbs

NTIAL ENERGY TOTAL BODY + LOWER

Normalized Potential Energy + Lower Limbs

AL BODY + LOWERLIMB RELATIVE ENE

Normalized Total Energy of Body + Lower Limbs

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