Analysis and Interpretation of Ground Reaction Forces in Normal Gait

TEA MARASOVIĆ, MOJMIL CECIĆ, VLASTA ZANCHI
Laboratory for Biomechanics and Control Systems
Faculty of Electrical Engineering, Mechanical Engineering and Naval Architecture
University of Split
Ruđera Boškovića bb, 21000 Split
CROATIA
tmarasov@fesb.hr, mcecic@fesb.hr, vzanchi@fesb.hr, http://labacs.hr

Abstract: - The locomotion biomechanics study provides very extensive and interesting material for investigating the physiological process involved and the neural mechanisms controlling the systems. Gait analysis – the systematic analysis of locomotion – is used today for pretretment assessment, surgical decision making, postoperative follow-up, and management of both adult and young patients. Over the past few decades, multiple advances in bioengineering technology have permited precise analysis of many specific gait characteristics, such as joint angles, angular velocities and angular accelerations (kinematic analysis), as well as ground reaction forces, joint forces, moments and powers (kinetic analysis); electromyographic activity and energy consumption. Sophisticated gait analysis equipment can generate a visible force vector on an oscilloscope screen and superimpose it simultaneously on a photograph of a gait subject. Visualizing ground reaction forces helps us understand the effect they are having on the body in walking. If we have a full kinematic description, accurate anthropometric measures, and the external forces, we can calculate the joint reaction forces and musle moments. This predicition, called the inverse solution, is a very powerful tool in gaining insight into the net summation of all muscle activity at each joint. In this paper, we present and discuss the results of a ground reaction force analysis, conducted using the data samples of 40 voluntary subjects from our faculty student and employee population, differing in age, sex, height and weight, measured by the AMTI force platform.

Key-Words: - biomechanics, kinetics, gait characteristics, ground reaction force, Pedotti diagram

1. Introduction

Kinetics by definition deals with those variables that are the cause of the specific walking or running pattern that we can observe or measure with cameras [1]. Kinetic (dynamic) movement quantities encompass forces and moments of force that are developed during movement between a body and its surroundings, as well as internal forces and moments, the mechanical power patterns (rate of generation or absorbtion by muscles, or rate of transfer between segments) or energy patterns (segment or total body) [5].

Transducers have been developed that can be implanted surgically to measure the force exerted by a muscle at the tendon. However, such techniques have only applications in animal experiments and even then to a limited extent. If a measurement procedure is to be noninvasive, the measurement of the remaining (internal) kinetic quantities of human movement is practically impossible. It therefore remains to calculate the joint reaction forces and muscle moments indirectly, using available kinematic description, accurate anthropometric measures and the external forces. This prediction is called an inverse solution and it is a very powerful tool in gaining insight into the net summation of all muscle activity at each joint.

Walking and running, the most natural forms of human locomotion, have often been the subject of ground reaction force measurements. As a part of the comprehensive research into healthy and pathological gait, Saunders [16] made the studies comparing the vertical ground reaction force signal components during stance phase in a healthy and unhealthy leg, specifically gait pathology resulting from a fused hip. The signals differed significantly because the remaining normal segments of the attacked extremity are only able to partially compensate for the loss of the hip joint function.

Hallet [10] researched the pathophysiology of posture and gait in individuals with celebellar ataxia. They used the AMTI platform and presented the signals of the ground reaction force vertical component of healthy subject and of a patient. The comparison revealed that there is a appreciable irregularity and weak formation of the initial phase of the force signal in the patient.

Other authors have focused their efforts on the research of characteristic biomechanical features (by means of EMG) of the locomotor system during normal gait and running, specifically the transition between these two modes of locomotion. In their article, Nilsson and colleagues [5, 13] presented the corresponding ground reaction force signals from two healthy subjects, at four different walking and running speeds, measured on the Kistler platform. Subjects wore athletic footwear. The results revealed the differences in ground reaction force signal waveforms in walking and running.

In gait, as well as running, ground reaction force signals reflect an increase in the movement speed through the even larger increase in peak values and a shortening of signal duration. Andriacchi [15] showed that there was a good correlation in some of the peaks of ground reaction forces and velocity of walking.

Music et al. [2, 3] proposed a method for measuring ground reaction forces during sit – to – stand motion using inertial sensors and a human body model. The proposed method fuses data from inertial sensors and the data from three – segment human body model using Extended Kalman filtering technique (EKF) which alleviates some of the drawbacks usually associated with inertial sensors. Dynamic human body model, incorporating shank, thigh and HAT (Head – Arms –Trunk) segments, was constructed based on principles of Lagrangian dynamics and the moments required in obtained model equations were calculated based on the EKF last best estimate and Newton – Euler inverse dynamic approach.

Data on the foot's center of pressure path during support phase in gait and running reveal interesting additional information. Besides vertical forces, Cavangh [13] researched the center of pressure paths during support in long-distance runners and discussed the results of the measured center of pressure paths made by two groups of runners differing significantly in their contact techniques. The first group was made of runners who realized the initial contact with the rear of the foot, and in the second group were those that made the initial contact with the mid – part of their foot. The implications of these measurements are interesting in connection with observing trauma caused by running – the study showed that the mid – foot strikers are exposed mostly to injuries and stress since they land relatively flat – footedly and the foot is not prepared for such loading.

The goal of this study was to make the comparison of the ground reaction force features between left and right leg during normal gait cycle. Single force platform was set up to measure barefoot, isolated steps. 40 student and employee voluntary subjects, differing in age, sex, height and weight, took part in the study. Data from this research were statistically processed and the results are presented in this paper.

2. Problem Formulation

The purpose of gait is to move the human body from one point to another. This represents a movement of the body center of mass along a horizontal trajectory in a certain direction and at a certain velocity. Since the human being does not use roller skates or other forms of wheeled transportation, there are certain unique features concerned with bi-pedal walking. The feet alternate in their ability to support the body and there are periods of changeover and propulsion that disrupt the smooth movement of the centre of mass in desired direction.

Each and everyone of us has a specific style of walking but within these different styles there are similar trends that can be classified in mechanical terms. Gait involves placing one foot forward followed by the next, then placing the same foot forward again and repeating the cycle of events. It is readily seen that gait is a cyclical activity and that one cycle can be analysed to represent a whole period of activity. However, it should be noted that each cycle within one individual suffers small variations due to distractions, different ground surfaces, etc.

In order to analyse the gait cycle in detail it is wortwhile to split the cycle into certain time periods. The most obvious approach is to represent periods of contact with the ground and periods when no contact exists. Taking one limb, every gait cycle encompasses two main phases: so called „stance“ phase, where the contact with the ground occurs, and the „swing“ phase during which the foot is not in contact with the ground. Within each of these phases there are certain time instances that are of great importance for the gait analysis.

The stance phase starts with the „heel strike“ (HS) at which point the contact with ground is made, Fig. 1. If the strike does not occur with the heel then it may be called ground contact. After the heel touches the ground, the foot is carefully controlled to come down towards the ground and provide a stable support base for the rest of the body. This is termed „foot flat“ (FF). Then the whole body rolls over the foot with the ankle acting as a pivot point and the hip joint is placed directly above the ankle joint at „midstance“ (MD). From that point onwards the purpose of lower limb is to propel the body centre of mass forward during the push off phase with an important time instant occuring when the heel loses contact with the ground at „heel rise“ (HR). As the foot leaves ground, it is usual for the toe to be the last point of contact, hence the end of the stance phase is often reffered to as the „toe off“ (TO). For normal subjects walking at their preferred speed on a level surface, the stance phase can take approximately 62% of the complete gait cycle (100%).

If there is a symmetrical gait then the left limb will perform the same events as the right limb. This means that the heel strike of the left foot should occur exactly halfway between heel strike of the right foot at the start of the gait cycle and heel strike of the right foot at the end of the cycle. If each foot is in contact with the ground for 62% of the cycle, there must be a 12% period of the cycle where both feet are in contact with the ground at the same time. This very important phase is the „double support period“ and represents the changeover and fine control for the smooth transition between limb support on the left and limb support on the right.

When walking velocity is increased the period of double support will be reduced. Eventually a velocity will be reached when there is no double support and this is the Olympic classification for „running“. As people start to run faster, the period of double support becomes negative and there is a flight phase.

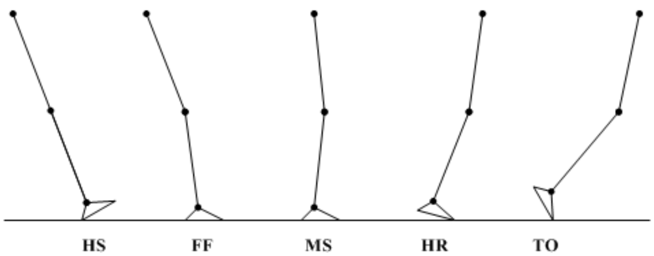


Fig. 1. The normal gait stance phase

The swing phase can be split into three phases rather than isolating distinct time instants. The first period involves the foot leaving the ground and accelerating forward and is logically called the acceleration phase. The ankle moves directly below hip joint during mid-swing and then the limb undergoes a deceleration phase ready for the next heel strike.

Although swing phase does not involve large forces of contact between the ground and the foot, there are two important items to be considered. The first consideration in the swing phase is that the limb must be shortened in vertical length so that the foot can clear the ground during the forward swing. The second parameter is that the masses of the lower limb will be subjected to accelerations and decelerations which will be represented as force actions requiring muscular effort at the hip and knee joints. As the swing phase is speeded up then the accelerations become larger and this becomes more demanding for musculature effort. To consider the limb movement during the swing phase of sprint runner, the kinematics or inertial parameters will be limiting factor. In some cases runners will produce more knee flexion so that the accelerations of the limb mass centers will be reduced by reducing the effective radius from the hip joint center of rotation.

The classical division of gait into heel strike, foot flat, midstance, heel off and toe off events can be inappropriate for some disabilities. For example, when analyzing the gait of a person with some sort of impairment, there may be no heel strike or heel off, the entire stance phase may be spent in foot flat. To avoid confusion and inaccuracies, another system of describing the phases of gait was developed [7, 8]. This system recognizes that there are three basic tasks during gait: weight acceptance, single limb support and limb advancement. To accomplish the task of weight acceptance there must be initial contact (phase 1) and then loading response (phase 2). During single limb support, midstance is the phase until the weight line is over the forefoot (phase 3), and then terminal stance occurs as the body weight moves anteriorly to the forefoot (phase 4). During stance, three „rockers“ are described [5], which contribute to the controlled forward fall of the body during gait. The heel rocker occurs during the loading response, the ankle rocker during midstance and the forefoot rocker during terminal stance. To accomplish the task of limb advancement there is preswing (phase 5), initial swing (phase 6), midswing (phase 7) and terminal swing (phase 8).

2.1. Measurement setup

To measure the force exerted by the body on an external body or load, a suitable force – measuring device is needed. Such a device, also called force transducer, gives electrical signal proportional to the applied force. There are many different kinds available: strain gauge, piezoelectric, piezoresistive, capacitive, etc. All these operate based on the principle that the applied force causes a certain amount of strain within the transducer. For the strain gauge type a calibrated metal plate or beam within the transducer undergoes a very small change (strain) in one of its dimensions. This mechanical deflections, usually a fraction of 1%, causes a change in resistances connected as a bridge circuit, resulting in an unbalance of voltages proportional to the force. Piezoelectric and piezoresistive types require minute deformations of the atomic structure within a block of special crystalline material. Quartz, for example, is a naturally found piezoelectrical material, and deformation of its crystalline structure changes the electrical characteristics such that the electrical charge across appropriate surfaces of the block is altered and can be translated via suitable electronics to signal proportional to the applied force. Piezoresistive types exhibit a change in resistance which, like the strain gauge, upsets the balance of a bridge circuit.

Ground reaction forces, acting on a foot during standing, walking or running, are traditionally measured by force plates. Force plate output data provides us with ground reaction force vector components: vertical load plus two shear loads acting along the force plate surface, that are usually resolved into anterior – posterior and medial – lateral directions. Refer to the Fig. xx for the coordinate system.

The fourth variable needed is the location of the center of pressure of the ground reaction vector. The foot is supported over a varying surface area with different pressures at each part. Even if we knew the individual pressures under every part of the foot, we would still be facing the costly problem of calculating the net effect of all these pressures as they change with time. Some attempts have been made to develop suitable pressure – measuring shoes, but they have been very expensive and are limited to vertical forces only. Therefore it is necessary to use force plate to obtain all the forces necessary for a complete inverse solution.

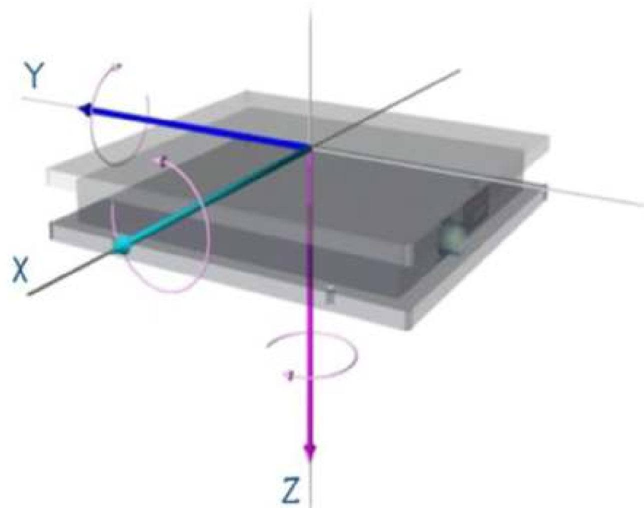


Fig. 2. Cartesian coordinate system of a force platform

The term center of pressure is often misinterpreted and even interchanged with the term center of gravity. Center of gravity of a body is the net location of the body's center of mass in the vertical direction. It is weighted average of the center of gravity of each body segment. It should be noted that the center of gravity is a displacement measure and that it is totally independent of the velocity and accelerations of the total body or its individual segments. The center of pressure is quite independent of the center of gravity. It is also a displacement measure and it marks the location of the vertical ground reaction force vector from the force platform surface. It is equal and opposite to a weighted average of the location of all downward (action) forces acting on the force plate. These forces are under the motor control of our ankle muscles. Thus the center of pressure is really the neuromuscular response to imbalances of the body's center of gravity. The major misuse of the center of pressure comes from researchers in the area of balance and posture, and they often refer to center of pressure as „sway", thereby inferring it to be the same as the center of gravity.

The tests described in this paper were conducted using a force plate with one centrally instrumented pillar which supports an upper flat. The dimensions of its contact surface are 50 × 50 cm and its mass is 11,4 kg. It was fixed into a wooden frame to decouple the influence of vibration from the unit mover and embedded into the floor to provide a more natural walk for the test subject. Each of the subjects was allowed to make two attempts per leg. The force data were sampled with 50 Hz. Fig. 3 shows the forces that act on the instrumented support.

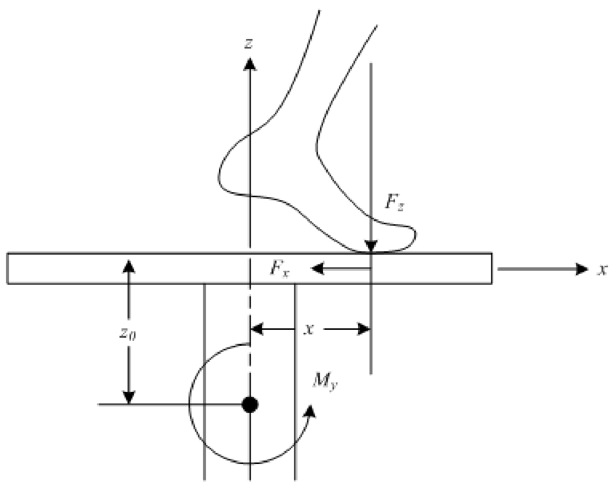


Fig. 3. Central support type force plate, showing the location of the center of pressure of the foot and the forces and moments involved.

These reaction forces are nothing more than the algebraic summation of all body segments mass – acceleration products:

$$F_x = \sum_{i=1}^{N} m_i \cdot a_{xi} \tag{1}$$

with m_i being the ith segment mass and a_{xi} being the ith segment center of mass acceleration in the x direction. Similarly, in the vertical direction:

$$F_z = \sum_{i=1}^{N} m_i \cdot (a_{zi} + g) \qquad (2)$$

where a_{zi} denotes ith segment center of mass acceleration in the z direction and g denotes the acceleration due to gravity.

With a person standing perfectly still on the platform, F_z is nothing more than the sum of all segment masses times g, which represents the body weight. The interpretation of the ground reaction forces as far as what individual segments are doing is virtually impossible. In the algebraic summation of mass – acceleration products there can be many cancellations. For example, F_z could remain constant while one arm is accelerating upward while the other has an equal and opposite downward acceleration. Thus to interpret the complete meaning of any reaction force, one would be forced to determine the accelerations of the center of mass of all segments and carry out the summations denoted with (1) and (2).

By summing all the moments acting about the central axis of the support, we can calculate the x coordinate of the center of pressure (CoP) of this ground reaction vector.

$$M_y - F_z \cdot x + F_x \cdot z_0 = 0 \qquad (3)$$

$$x = \frac{F_x \cdot z_0 + M_y}{F_z} \qquad (4)$$

where M_y represents a bending moment about axis of rotation of support and z_0 denotes the distance from support axis to force plate surface.

The reference point of origin is bellow the floor surface, hence the effect of F_x on M_y is negligible. Since F_x, F_z and M_y change continuously over time, x can be calculated to show how the CoP moves across the force plate. However, caution is required when F_z is very low (< 2% of body weight) because, during this time period, a small error in F_z causes a large percentage error in x, as calculated by (4).

Similarily, the y coordinate of the center of presure of the ground reaction force vector can be obtained by through the following equations:

$$y = \frac{F_y \cdot z_0 + M_x}{F_z} \qquad (5)$$

3. Results and discussion

Typical force plate data are shown in Fig. 4. plotted against time for a subject walking at a normal speed.

The vertical component of ground reaction force F_z, shown on Fig. 4, is the largest component and accounts for the acceleration of the body's center of mass in vertical direction during walking. The force curve is sometimes called the „M curve“, due to its particular shape, which resembles that letter.

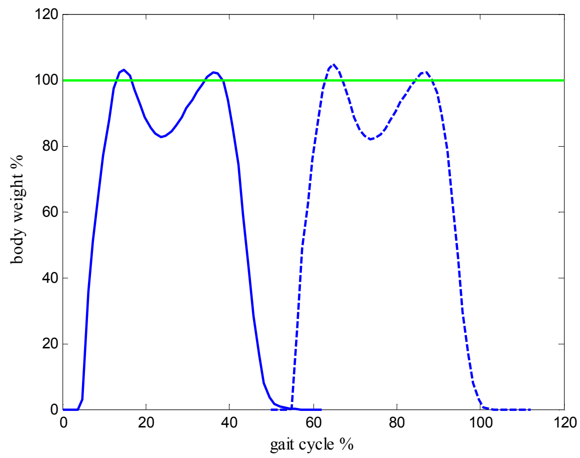


Fig. 4. Mean vertical (F_z) ground reaction force component waveform exerted by right and left (dotted line) foot in walking at normal speed.

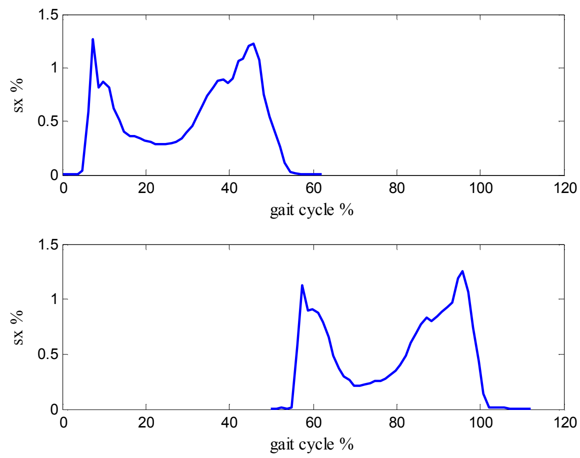


Fig. 5. Vertical ground reaction force measurement standard deviation for a) right and b) left foot.

Let us consider the individual components of the stance phase: double and single stance for one leg, Fig.

4. At the instant of heel contact, there is zero force between the foot and the floor due to the fact that the heel strike is defined the moment when contact is generated but there is no force production at that time instant. For a very small increment of time after a heel strike, the force will start to increase rapidly as the body begins to be supported by the foot.

Soon after, full body weight will quickly be generated between the foot and the floor. Since the center of mass is moving downwards and under some deceleration, the inertial force has to be added to gravitational force, which means that the vertical force applied to the foot exceeds body weight and peaks at the value of approximately 107 % of body weight, as it can be seen on Fig. 4.

As the gait cycle progresses through single stance, it could seem unusual that the ground reaction force should be less than the body weight during the single stance when only one foot is on the ground. This is made clear if we know that, at this stage of gait cycle, the center of mass experiences a downwards acceleration, which will give an upwards inertial force that will reduce the ground reaction force to 85 % of body weight.

To further understand these fluctuations of ground reaction force around body weight, the following should be appreciated: if the human body is resting in a standing position, then the ground reaction force will be equal to human body weight. But if the center of mass of the whole body is in motion and under some acceleration or deceleration, the produced inertial force has to be added to or subtracted from body weight, depending of inertial force direction. It is known that the vertical acceleration can be up to 20% of gravitational acceleration upwards and downwards, which means that the vertical ground reaction force will be 100% ± 20% of body weight.

In the final stage of the stance phase (push off) the propulsive action generates forces greater than body weight. Second peak appears near the middle of gait cycle and reaches 105 % of body weight. As the limb aims for the toe off, the force rapidly reaches zero value at the end of stance phase.

The anterior – posterior reaction force F_x represents the horizontal force exerted by the force plate on the foot (Fig. 6). F_x acts in a direction of human walking: upwards and backwards and is smaller in magnitude than vertical ground reaction force component, typically reaching around 20% of body weight. At first it is a braking force to midstance, in order to decelarate the body center of mass, and then it is followed by propulsion.

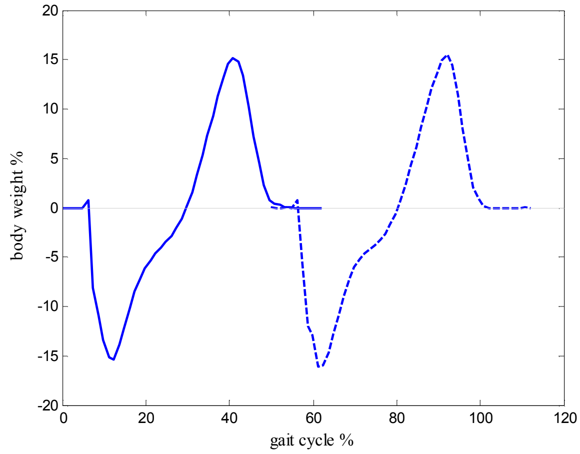


Fig. 6. Mean anteroposterior (F_x) ground reaction force component waveform exerted by right and left (dotted line) foot in walking at normal speed.

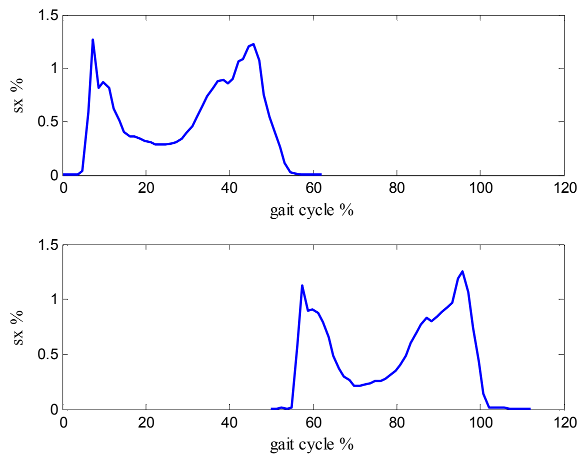


Fig. 7. Horizontal ground reaction force measurement standard deviation for a) right and b) left foot.

At heel strike and toe off, zero force occurs. In the first stage of double stance, a forward directed force is generated. This is calles „claw back“ because, due to the initial parameteres, the foot claws back just before contact with the ground is made. This is a sort of a „shock“, which can also be seen on a vertical ground reaction force component if sampling frequency is high enough (around 200 Hz). On Fig. 4. this rapid built of vertical force isn't visible due to a small sampling frequency, which was 50 Hz.

After the „claw back“ disappears, a backwards force must be applied to the foot by the ground in order to decelerate the body center of mass as it approaches a single stance. This is shown as a rapid increase of backward force. Afterwards, the force is falling, approaching zero value in the middle of single stance.

In the latter half of the single stance phase, to accelerate the body center of mass in a forwards direction, the ground applies a forward force on the foot. This is shown as a gradual, and then rapid, build up, followed by a rapid reduction in the force as toe off approaches.

It can be observed that positive and negative forces are almost symmetrical, which could be explained by the change of impulse. The area under any segment of a force curve represents the impuls (time integral of the force). This impuls can also be defined as a change of the momentum ($m \cdot \Delta v$). If the individual is walking at a constant speed, there is no change in the momentum ($\Delta v = 0$), and the total impulse in the anterior – posterior direction should result to zero. That means that braking impulse is approximately equal to the propulsion impulse in balanced gait. This is an important fact for the analysis and diagnosis of pathological gait patterns.

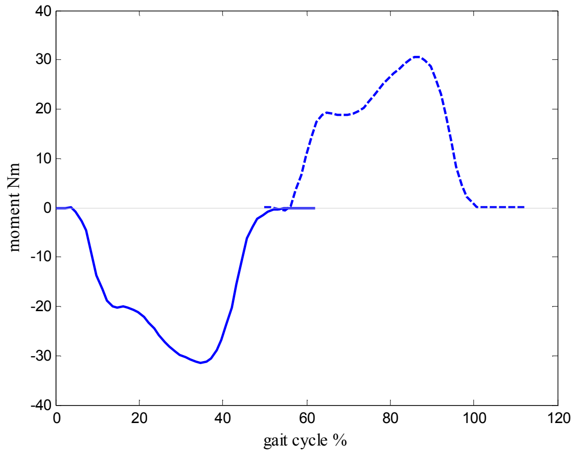


Fig. 8. Mean moment of force about x – axis (M_x) waveform exerted by right and left (dotted line) foot in walking at normal speed.

The moment about the x – axis (Fig. 8) is a consequence of the swaying motion performed by the subject, whereas the moment about the y – axis (Fig. 9) is due to the shifting of weight when subject is standing on one foot. The peaks in both, positive and negative direction refer to the time at which one foot was placed on the platform. The moment about the z – axis is very small, as can be seen in Fig. 10. This is because there is no real way for the subject to rotate the force about the z – axis during normal gait. The subject would have to apply a twisting motion of the foot to create any significant result.

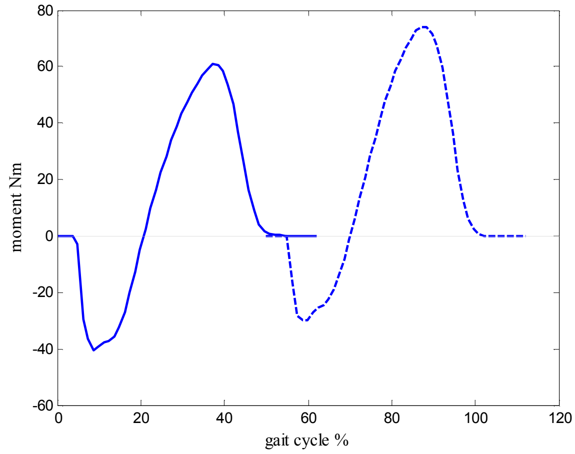


Fig. 9. Mean moment of force about y – axis (M_y) waveform exerted by right and left (dotted line) foot in walking at normal speed.

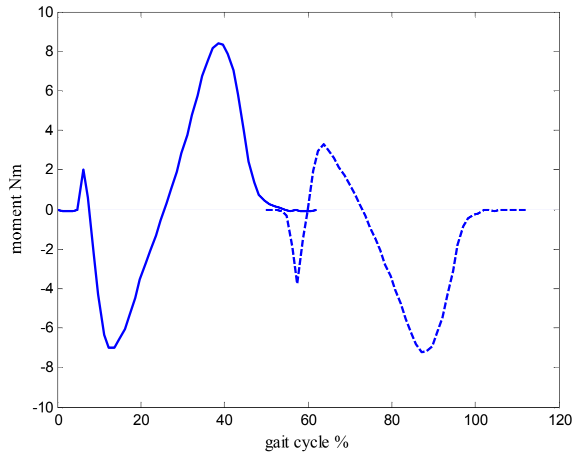
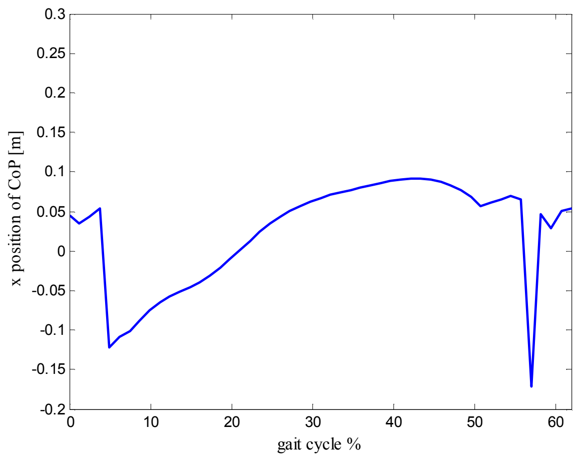


Fig. 10. Mean moment of force about z – axis (M_z) wareform exerted by right and left (dotted line) foot in walking at normal speed.



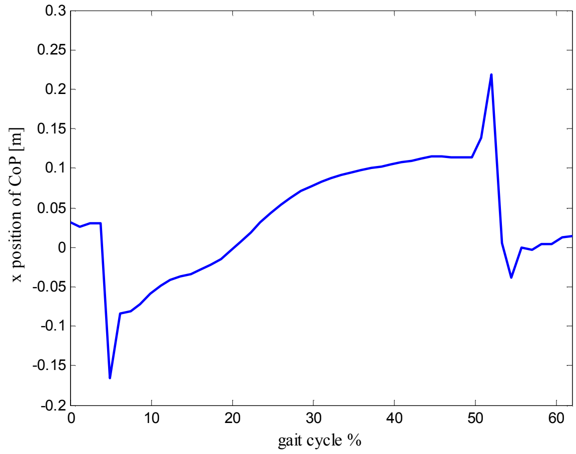


Fig. 11. Mean center of pressure positions along the *x* – axis throughout stance phase of gait cycle for (a) right and (b) left foot.

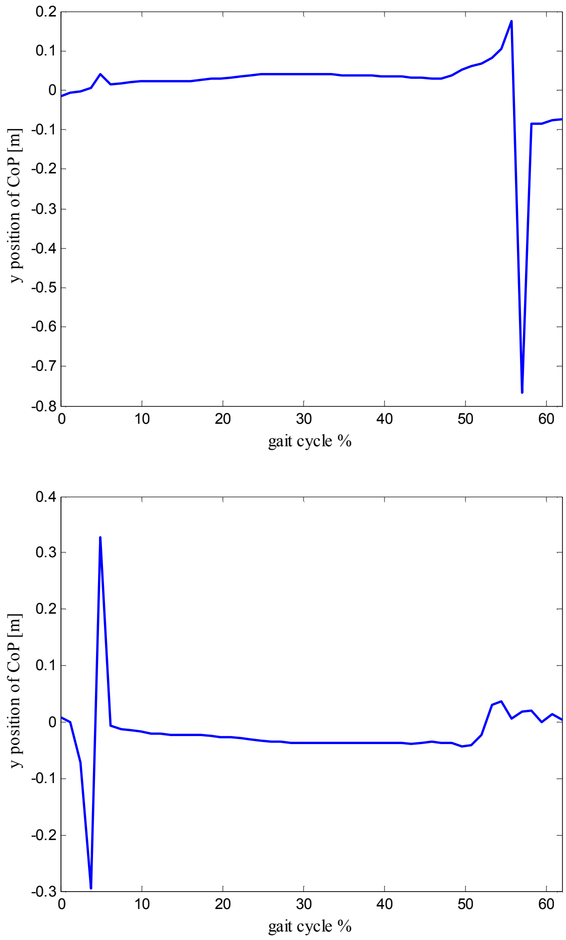


Fig. 12. Mean center of pressure positions along the *y* – axis throughout stance phase of gait cycle for (a) right and (b) left foot.

As it was discussed in the previous few paragraphes, the vertical and horizontal component of the ground reaction force are continouosly changing, thereby changing the direction and magnitude of the resultant ground reaction force. As a subject walks, its center of pressure moves forward, starting at a heel, assuming that the initial contact was made by the heel, and then progressing towards the ball and the toe. The position of the center of pressure relative to the foot cannot be obtained from the force plate data themselves. First the foot's location relative to the midline of the plate has to be determined.

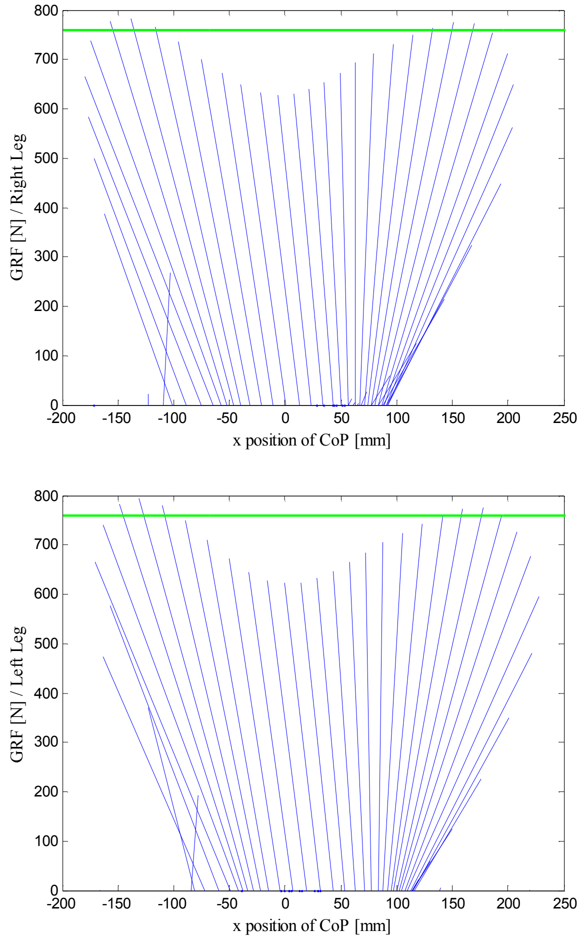


Fig. 13. Pedotti diagrams in sagittal plane: ground reaction force combined with *x* positions of center of pressure, (a) right foot, (b) left foot.

If we combine vertical and horizontal component of the ground reaction force with center of pressure *x* coordinates we can obtain a graphic representation of complete spatiotemporal sequence of the ground reaction evolution, during stance phase of gait cycle in the sagittal plane. In this vector diagram, also known as Pedotti [14] or Butterfly diagram, each vector presents the projection of ground reaction force, with its magnitude, inclination and point of application in uniformly discretized time instants, on sagittal plane (in direction of progression). The same principal, using the center of pressure y coordinates can be applied to the frontal plane.

The diagrams, shown in Fig. 13, have a shape of a butterfly wing. At normal gait speed the envelope of the vectors presents a typical pattern characterized by two maxima and one minimum. On the right, in the first stages of stance, when the foot touches the ground, except for the first few vectors that absorbe the shock due to the heel strike, all other vectors are inclined contrary to the direction of movement. In the later stages, the leg pushes the entire body up and forward and hence the vector inclination faces the movement. The velocity with which the center of pressure moves on the ground is immediatly readable as the density of the vectors. As we can see, the center of pressure is moving monotonously in the direction of progression and is showing an evident plateau in the later stages of the stance which results in correspondent closeness of the vectors.

4. Conclusion

The ground reaction forces, as measured by a force platform, reflect the net vertical and shear forces acting on the surface of platform. As such, they are an algebraic summation of the mass – acceleration products of all body segments while the foot is in contact with the platform. The shape of these reaction forces is typical of what is reported in the literature.

The vertical force F_z has a characteristic double hump. The first hump appears at heel contact, showing a rapid rise to a value in excess of body weight as full weight bearing takes place and the body's downward velocity is being arrested. Then, as the knee flexes during midstance, the plate is partially unloaded and F_z drops below body weight. At push-off the plantarflexors are active, causing a second force peak greater than body weight, which demonstrates that the body's centre of mass is being accelerated upwards to increase its upward velocity . Finally the weight drops to zero as the opposite limb takes up the body weight. Unfortunately, the vertical load curve has been found to be unreliable as a clinical measure. This is largely due to its sensitivity to any action or reaction altering the ground reaction force vector, such as arm lifting, which can diminish the peak component to less than body weight. Children are reported to have somewhat lower average vertical and shear peak forces compared to adults.

The horizontal force has a negative phase during the first half of stance, indicating a backward horizontal friction force acting between the ground and the foot. If it wasn't for this this force, the foot would slide forward, as it happens when one is walking on icy or slippery surfaces. Near midstance F_x gets positive, which signifies that the force plate reaction is acting forward as the muscle forces (mainly plantarflexor) cause the foot to push back agains the plate. Both shear and center of pressure measurements are also affected by position and motion of all body segments such as head, arms, trunk, pelvis and legs.

The Pedotti diagram further reveals that the largest magnitudes of force occur at the heel and forefoot, supporting the notion that the foot in fact acts as an arch. From all the tests performed, in agreement with the vector diagrams presented here, one can conclude that there is a great similarity in records of left and right leg at certain gait velocity, which implies a symmetry of the stride in normal population.

Current and future directions in gait analysis will include more sophisticated tools for analysis and interpretation of data such as pattern analysis, neural networks and artificial intelligence.

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ACKNOWLEDGEMENT: This work is sponsored by Republic of Croatia Ministry of Science, Education and Technology, Project No. 023-0232006-1655 „Biomechanics of Human Motion, Control and Rehabilitation“