Advances in Science, Technology and Engineering Systems Journal Vol. 6, No. 2, 58-64 (2021)

www.astesj.com

Special Issue on Multidisciplinary Sciences and Engineering

ASTESJ

ISSN: 2415-6698

Methodology for Calculating Shock Loads on the Human Foot

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A R T I C L E I N F O

Article history:
Received: 24 December, 2020
Accepted: 14 February, 2021
Online: 10 March, 2021

Keywords:
Foot
Musculoskeletal System
Step Cycle
Locomotion
Biomechanical Parameters
Support
Elastic Characteristics
Coefficients of Capacity
Shock Load

A B S T R A C T

The leading place among diseases of the musculoskeletal system is occupied by various feet deformations. Clinical movement analysis and posturological examination are required to objectively assess the distribution for load caused by the weight of human body on the feet and its locomotion effect. In normal conditions, the foot is exposed to elastic deformations. When analyzing the foot loads, it`s necessary to consider shock loads as one of dynamic load types. The foot is the first to perceive the shock impulse by support reaction, and the further nature for interaction with the environment directly depends on its functional capabilities. However, the foot supporting properties haven`t been fully researched. The purpose for this research is to increase the accuracy of estimating the human foot biomechanical parameters, by assessing the dynamic impact, namely short-term shock loads by step cycle relevant phases. This goal is solved by developing a method of static-dynamic load analysis, which allows to estimate dynamic and shock loads on foot and is reduced to determining the capacity coefficients, dynamic and shock loads. In the course of studies, conducted in this research, it was found that the maximum contact per unit time has front section (repulsion phase), then - the rear section (landing phase) and the smallest - the foot middle section (rolling phase), the greater speed and length step – so the greater shock loads coefficient, and their peak falls on the front and rear sections. The practical significance of the obtained results is to improve the existing methods of researching biomechanical parameters by comprehensively assessing by standing and gait features, foot step cycle and support properties.

1. Introduction

The current stage of development for instrument making actualizes the applied aspect for human locomotions analysis. Diagnosis of posture wouldn`t be complete without measuring and assessing the state of support-spring foot properties. Clinical analysis for movement and posturological examination (body position examination) are required to objectively assess the effect by foot load distribution on locomotion. In modern biomedical engineering, a fairly promising area is clinical analysis of motor activity - the research of various pathologies of gait and main rack, using biomechanics methods. There is no doubt about the importance of assessing the functional state by musculoskeletal system in patients with orthopedic and neurological profile [1].

In recent years, the number of diseases, injuries and pathologies of the musculoskeletal system (MSS) is growing steadily, which significantly affects the quality of life. Analysis of human gait is subject by many scientific studies. Various diseases and injuries of supporting organs are often accompanied by serious functional disorders of musculoskeletal system, decreased muscle strength and tone, loss of ability to normal movements, which ultimately leads to disability and subsequent disability [2, 3]. The foot is structural segment in musculoskeletal system, which provides it`s stomato-motor function, and is integral morphofunctional object that human motor function depends [2]. By exteroreceptors located on plantar surface, and it collects information about body mass fluctuations and directs it to the central nervous system (CNS), which will coordinate postural stability [4].

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https://dx.doi.org/10.25046/aj060208

The foot - is the first, most loaded part of musculoskeletal system, which makes contact with the support, redistributes the reaction force to the higher segments for musculoskeletal system and performs an important spring function, provides stability of the lower limb and adhesion to support surface [5]. There are three main functions of foot:

- spring - the ability to elastic stratification under the action of load and restore its shape after removal of the latter;
- balancing - participation in the regulation of posture and positional activity when standing and walking;
- push (repulsive) - the transfer of acceleration of the total center of mass (TCM) for all body during locomotion [3].

During life, the functional parameters of foot change to one degree or another. First of all, they relate to its spring and repulsive functions. In the case by violations of symmetry for load distribution on foot, as well as by reducing its elastic characteristics - reduced damping properties, which leads to increased shock loads and increased vibration effects on the entire MSS [6]. Currently, longitudinal and transverse flattening of the feet, which can be both independent and in combination with other deformities, is one of the most common orthopedic diseases. According to the WHO, 75% of people have some pathological changes in the feet, which the most common is flat feet (changes in the shape of foot, that is characterized by a decrease in the height of its transverse and longitudinal arches). This deformity is the cause of many serious MSS diseases, which often lead to disability [2, 3].

Analysis of relevant literature sources [6-12] shows the current state of injuries biomechanics and foot deformities. In particular, a significant number of scientific publications aimed at researching the condition by lower extremities of different groups for population and identifying the relationship between foot deformity and other structural disorders of MSS were considered. Researchers note that the problem of early diagnosis of injuries and diseases for feet is relevant in the choice of prevention methods, treatment, orthosis and their effectiveness evaluation. Given that foot bears the main load, the violation of its functions is reflected in entire musculoskeletal system functioning and can lead, in the future, to a number of chronic diseases. The research of human foot in the dynamics (with gradual or sharp variable load on it) is a key factor in the diagnosis for its functional state, which allows you to identify abnormalities and determine the necessary set of treatment measures [6-12]. There are methods for determining the biomechanical properties for foot soft tissues, based on the assessment changes in the pressure values applied from the outside, sufficient to block the arterial vascular bed of tissues, determined by discoloration for skin surface or disappearance of arterioles in blood vessels. However, these methods have a number of limitations, in particular, can`t be used in the assessment of gait. During walking, there are additional short-term shock loads of the anterior and posterior shock, in the damping of which may involve various biological media, in particular, in addition to musculoskeletal, fluid blood-lymphatic media play a significant role [6]. Also in scientific publications described the types and variants of gait at various deformations, scoliosis and number of skeleton congenital deformations. The authors of some publications point to the imbalance and lack of synchrony in the muscles for lower leg and foot, changes in TCM, instability when walking and rapidly progressing pain in various pathologies [11, 12]. However, among the variety of scientific approaches to addressing this issue, the research for distribution the foot load during the full step cycle in dynamics, not fully studied. Known methods don`t allow to diagnose functional changes in the foot that occur when the load changes, as well as to determine the individual physiological characteristics of lower extremities, which limits the use of such methods. In addition, in domestic and foreign literature not enough attention is paid to quantitative assessment for parameters of the foot elasticity, while the latter analysis will improve the MSS accuracy diagnosis.

At overloads of arch supporting systems - functions of foot are broken, the motor stereotype as a whole is distorted, there are undesirable redistributions of forces and overloads which are transferred to other parts of MSS therefore there are pathologies. In such cases, the foot works not as an elastic-elastic system, but as an elastic-plastic system, with its inherent residual deformation. The most common cause of the latter is overload associated with functional impairment of foot arches, i.e. recompensation of arches is expressed in a particular injury [6]. Violation for arches formation to the feet occupies 26.4% of all orthopedic pathology and up to 81.5% for all deformities of lower extremities in children [12]. Researches show that in children`s these disorders can be partially or completely corrected, while in adulthood - these disorders are extremely difficult to correct, sometimes impossible. It has also been proven that the absence of physiological arches and disturbances for ankle axis joint leads to development of pathological processes in the large joints of lower extremities and spine, and is often the cause of pain [5, 6]. With foot flat-valgus deformity, in addition to muscle weakness and ligaments, the shape and foot bones ratio are disturbed, the reference vector is displaced laterally (to foot outer edge). This changes the nature of gait, is the cause of increasing the dynamic load on the entire musculoskeletal system [2].

Based on this, various foot deformities, in particular flat-valgus, should be considered as primary link in unstable gait formation, which affects the violation of MSS, and leads to various lesions of the latter. The vast majority of modern tool systems are physical rather than linear, which complicates the search for solutions on an elementary basis and sometimes makes it impossible. It`s in this statement that modern instrumental analysis develops. For linear systems, the minimum change in the initial state - causes the corresponding changes by its final state. Otherwise, for nonlinear systems, small differences in the initial conditions lead to unpredictable changes in final state. For example, a person undergoing static analysis doesn`t know how much and how to change their motor behavior during dynamic analysis [4]. In this regard, morphofunctional diagnosis of the foot is an essential element in the prevention of MSS number disorders, and the applied aspect for analysis of human locomotion, in modern biomedical engineering, is very relevant and promising.

2. Materials and methods

2.1. Formulation of the problem

Clinical analysis of movement and posturological examination are required to objectively assess the distribution of load caused by the weight of human body on feet and its effect on locomotion. There is baropodometry method, which allows you to objectify the research of biomechanical parameters of foot, taking into account the static and dynamic components. This allows for more in-depth research of locomotions and to detect a certain frequency of identical or pressure similar images, which don`t fully correspond to the known metatarsal formula [1-5]. Thus, the research for biomechanics of the lower extremities in norm and at various deformations is necessary and very promising in terms of assessing the state by MSS and forecasting its dynamics.

In order to increase the accuracy for research of biomechanical parameters of foot, it`s necessary to take into account the influence of dynamic, namely short-term shock loads caused by body weight on the corresponding phases by the step cycle.

2.2. The foot impact load calculating method

The article task - to identify the initial stages of injuries and foot deformations, is solved by developing a method of calculating the dynamic load taking into account the short-term foot impact.

The foot ability to withstand various loads is due not only to biomechanical perfection, but also the properties of its constituent tissues [2]. Normally, due to vaulted structure of the foot and its spring function, up to 70% of the acceleration is damped and amortized. In normal development of the musculoskeletal system, the load is distributed as follows: through the body of the heel talus, navicular and cuneiform bone, then on the heads of I-III metatarsals, forming an external longitudinal arc (As shown in Figure 1) [3]. The foot biomechanics and its functions in different phases of the step cycle are different. Mitigation of the inertial load during walking and running is carried out by articular ligament difficult complex, which connects the 26 main foot bones, where there are 5 longitudinal bones and the transverse arch. The heel, talus and metatarsal and metatarsal bones form a kind of arch - a spring that can shrink and straighten under the loads action. Body weight is normally evenly distributed on the front and back of the foot. These divisions are connected in a single kinematic chain by inter-articular ligaments, as well as a strong elastic tendon - plantar aponeurosis, which, like a spring, returns to normal position spread out under the load of the foot arch [1-5].

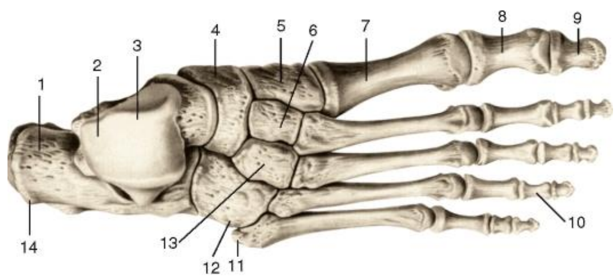

Figure 1: The human foot structure [3]

1 - heel bone; 2 - talus block; 3 - talus; 4 - navicular bone; 5 - medial wedge-shaped bone; 6 - intermediate wedge-shaped bone; 7 - I metatarsal bone; 8 - proximal phalanx; 9 - distal (nail) phalanx; 10 - middle phalanx; 11 - hump V of the metatarsal bone; 12 - cuboid bone; 13 - lateral wedge-shaped bone; 14 – humerus

The gait main functional unit - cycle of the step, that time from the beginning of limb contact with the support to the next same contact with the same limb. The normal locomotor cycle consists of two bipolar and two portable phases. The average step cycle time, at normal speed, is close to 1*s*. The step cycle, for each limb, consists of three main periods: the support period, the transfer period and the double support period. Research in the field of biomechanics has shown that the forces applied during movement have vertical and horizontal components. Gait is a complex cyclic movement associated with the repulsion of the body from the support surface and its movement in space. The efforts made are dynamic. Characteristic of normal gait is the constant storage of support on one or two limbs [2].

The basis for any locomotor act is the support interactions, i.e. short-term contact of a MSS certain link with the support, as a result of which there are forces capable of changing the motion center of the body's TCM. Support interactions have all the physical features for shock loads (short duration, significant increase in the modulus of force, etc.), so such interactions can be considered as shock [6].

The algorithm for calculating shock loads is:

- Step cycle registration;
- Determination of biomechanical parameters for step cycle - load values, speed, contact areas of support surfaces, half-step length, etc.;
- Determination for dynamic capacity coefficient of the lower extremities;
- Determination of the dynamic load factor;
- Determination for impact load factor.

2.3. Determination of step cycle biomechanical parameters

During the movement in each step cycle, there are two phases of double and single support. The moment of time t_{fi} of the maximum vertical pressure $P_{v.max}$ s called the forward shock. The time t_{bi} of the rear shock always coincides with the maximum $P_{l.max}$ of longitudinal forces that move the body forward, at this time t_{bi}, the maxima $(P_{v.max} + P_{l.max})$ of the vertical and horizontal forces are summed. Between the moments of time $\Delta t=(t_{bi} - t_{fi})$ of the front and rear shock of the lower extremity there is a damping failure $\Delta P_f=P_{v.min}$, which corresponds for minimum vertical pressure. When the overall center of gravity rises higher, the pressure on the support is directed forward, replaced by the pressure directed backwards, i.e. the braking is replaced by a push. In healthy people, the step reference period is $t_{rp} = 0{,}64 \pm 0{,}016\ s$, the transfer period $t_{tp} = 0{,}36 \pm 0{,}014\ s$, the two-support period $t_{trp} = 0{,}135 \pm 0{,}010\ s$. The duration of support time on the right and left foot usually differs by $\pm 5\%$ and is [2]:

$$\Delta t_{tp} = (24—29)\% \cdot t_k \tag{1}$$

where t_k - step time.

Given the different pace, these rates are estimated as a percentage for entire step cycle: the period of rolling through the foot is equal to [2]:

$$\Delta t_r = (45—51)\% \cdot t_k \tag{2}$$

The period for the lower limb over the support transfer [2]:

$$\Delta t_{tp} = (31—41)\%\cdot t_k \tag{3}$$

Bipolar step period [2]:

$$\Delta t_{trp} = (8—24)\%\cdot t_k \tag{4}$$

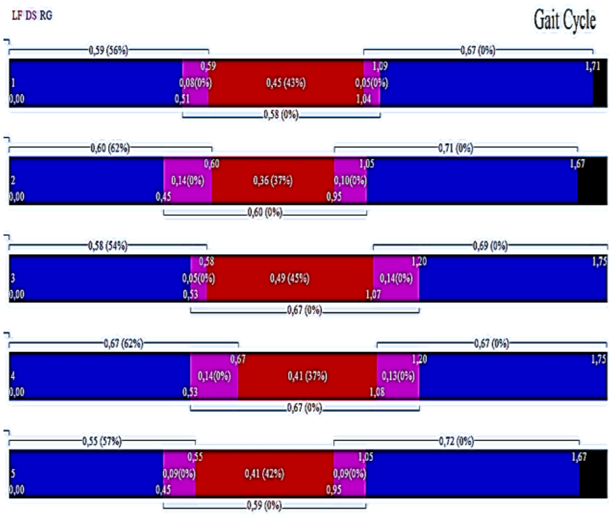

Figure 2: The human foot structure [2]

In fig. 2 [2] presents step cycle graphical representation. Red indicates the period of support on left lower limb, blue - on the right, pink indicates the period of double support (two limbs), black characterizes the contact lack. After analyzing the step cycle, we can conclude that the period of transfer for one limb is equal to single support period on the opposite limb [2].

In order to identify the leading limb, we introduce the dynamic capacity coefficient [2]:

$$K_{dS} = P\cdot t \tag{5}$$

where P – foot pressure on the support surface with a single support (kg/m^2),

t – the duration of the support period (s).

In previous researches [2] was determined the dynamic coefficient q, which is calculated as ratio of force transmitted to the resistance during locomotion to body weight m:

$$q = \frac{P}{m} \tag{6}$$

The following values for dynamic coefficient are calculated and determined q:

- for slow walking $q \leq 1$
- for fast walking $q \leq 1{,}5$
- for running $q \leq 1{,}8$

In the first iteration, the dynamic coefficient q will be equal to one (the beginning of the movement), so the pressure for foot P will be proportional to the body weight m. Given the latter - the maximum value of the coefficient for dynamic capacity will be the longest period of single support. In order to identify the leading limb, we write the ratio of the coefficients K_{ds} for the left and right lower extremities, and obtain [2]:

$$\frac{K_{dSL}}{K_{dSR}} = \frac{t_L}{t_R} \tag{7}$$

where $K_{dSl/r}$ – dynamic capacity coefficients for the left and right lower extremities, respectively.

Twenty-five children aged 3 to 17, with an average body weight of 45 kg, were selected as the subjects for this research. The research tool base was a baropodometric platform, measuring 0,4×1,8×0,02 m, (As shown in Figure 3) with appropriate software.

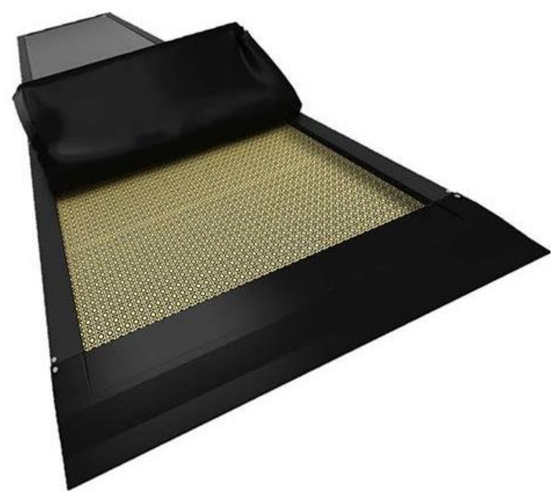

Figure 3: Digital baropodometric platform [3]

The main technical characteristics of the system are given in the table 1 [3].

Table 1: Biometric system technical parameters

Parameters	
Measurement error	± 5%
Scan type	Matrix
Resolution, dpi	x, y=9,6 z=16
Pressure on a point, kg/m^2	max 150·10^4
Measurement frequency, staff/s	60
Sensor type	High-resistance sensor with active matrix
Number of sensors on 1 m^2	400
Sensor dimensions, m	0,025
Power supply	AC input: 100-240V - 1A, 50-60Hz, DC output: 12V – 2,5A
Compatibility with operating systems	Windows 10-64 bit
Interface	USB 3,0

The method of baropodometry was used in the research. The platform consists of 4 active modules, measuring 0.4x0.4 *m* and 4 passive. 6400 sensors are installed in each active module (the total number of sensors on the platform is 25600) [1]. The platform works in two modes: static and dynamic.

Baropodometry of static position determines the distribution of load zones, outlines the perimeter of the support polygon, fixes the centers for foot position and the projection of TCM and its displacement, calculates the percentage of support surface and pressure force, including limb overload or pelvic rotation. Static analysis is considered as a geometric model that relates basic biomechanical parameters, correlating with the information obtained from the morphology of the sole (podocontourmetry), and with the reflection for foot pressure obtained in the dynamic analysis. Quantitative values of pressure imprints, detected after the research and presented in the section of static analysis, allow you to notice any possible asymmetry or deviation from the physiological state. The most important of these are divided into "load values" (correlated mainly with pressure information) and "offset values" (correlated mainly with spatial information) - they are interrelated [6].

Baropodometry in the dynamics shows how the pressure is distributed during the rolling to each foot. The point of landing, contact and shock normally have a clear sequence, speed and strength. The graphical representation of the movement can clearly track the stability of the joints, lateral or medial deviations of movement. In the process of analysis, cycles of steps with time characteristics of mono-support and double support are recorded. Elongation of the foot with dynamic support, its expansion in the anterior part during movement are determined. The full step cycle is analyzed for three prints, and a half-step for two. Having a complete step cycle is always a more informative parameter than single prints. When walking, there can also be a supporting mid-foot polygon (As shown in Figure 4), i.e. the moment at which the body transfers the load from one limb to another during the double support. This parameter is crucial in the kinematic reconstruction of stability disorders, which can be caused by incorrect contact, because it allows to detect the interaction of possible stability disorders [4].

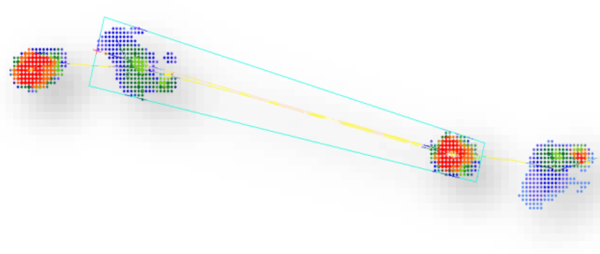

Figure 4: Supporting mid-foot polygon [4]

The research of the load distribution on foot plantar surface. According to formula above, determine the dynamic capacity coefficient *q* (6). After the corresponding calculations it was obtained: for 40% of the group the left lower extremity has the maximum period of support, in the rest of the group - the right. Further calculations were performed for a larger group, i.e. on the left support.

3. Results and discussions

In order to analyze the distribution of dynamic load on the support surface, let`s introduce the coefficient of dynamic load K_{dl} [2]:

$$K_{dl} = P_i \cdot S_i \cdot t_i \tag{8}$$

where P_i – the value of the load on the i-th characteristic of the cycle for foot step (the period of push, roll and landing), kg/m^2;

S_i - characteristics area, m^2;

t_i - duration of the characteristic, *s*.

Impact is a load that is applied in a very short period of time. For example, the shock load occurs when one body falls on another or when the pressure between the bodies in question changes rapidly. It should be borne in mind that if considered a body with a certain mass m, such as a weight of 20 *kg*, the weight of the weight will not change before, after or at the time of fall. This statement is true only when it comes to gravitational mass, but a thorough world research of the phenomena shows that any body also has an inert mass. Researches show that the inertial mass is equal to the gravitational mass, and when it comes to shock loads, they are created not by gravitational but by inert mass. From the general course of physics, the term "load" is not used, but the concept of "force" is used. In this case, all forces can be divided into external and internal. In the course of theoretical mechanics, the theory of resistance of materials, the theory of elasticity, the theory of strength, etc. external forces acting on a particular structure are considered as loads, and internal forces - as stress. It`s assumed that the sum of external forces is equal to the sum of internal forces, this ultimately allows you to make an equilibrium equation for the selected system [2, 6]. In this research considered the human body with mass *m*, as a concentrated load Q_{st}, which according to Newton's second law is calculated:

$$Q_{st} = m \cdot g \tag{9}$$

where *m* – body weight, *kg*;

g – acceleration of gravity.

Due to the support interaction of body weight with foot, in the latter there are certain ratios of elastic forces - the internal force field, which counteracts the occurrence of plastic deformations by support caused for body weight. If the support conditions change (for example, when changing posture), then the configuration for the force field of elastic forces in the considered system changes accordingly. In this case, if a person doesn`t move, i.e. the speed of its movement relative to the considered frame of reference is “0”, then his body still creates a load - static. Here, a human foot is considered as a support that acts with force equal in value to the reference force of the human body *F* (load caused by the mass of the body applied to the support) and opposite in direction, such a force is called the support reaction R_d. When $R_d=F$ – the system is in equilibrium. After the start of motion with a certain acceleration *a* there is a vertical force of inertia directed opposite to the acceleration. Accordingly, if the inertia force is directed downwards, the load caused by the body mass on the resistance increases, because under the influence of acceleration, the static load Q_{st} is replaced by dynamic Q_d:

$$Q_d = Q_{st} \cdot \frac{a}{g} \tag{10}$$

Relation *a/g* is a measure of mechanical overload, which determines the change in physical condition of the body. If the load is applied instantly - the reference force will be the maximum value, because the body is mass *m*, at a certain point in time *t* will reach a fairly high speed *v*, therefore, the calculations must take into account the gravitational, inert mass of the body and the inert mass of the system (body + support). Therefore, when we consider the types of loads on the human foot, it is necessary to consider shock loads as one of dynamic loads types, which differ from static ones in that the maximum stresses should take into account the forces of inertia.

If we consider the body of person with mass *m* as a physical body that creates a load, and his foot - as a body that receives this load with the occurrence of certain stresses, then the interaction of the body with the foot - the speed of gravity of both bodies, according to accepted frame of reference, do not change. This allows us to consider the loads and stresses caused by them as those that are due only to the gravitational interaction. The foot as a supporting structure is the first to perceive the shock pulse p of the reference reaction, and the further nature of the interaction with the environment directly depends on its functional capabilities.

$$p = m \cdot v \tag{11}$$

where v – speed of movement n-th phase of the step cycle, *m/s*.

In this research, we consider a rectilinear (translational) motion, i.e. one for the description of which it`s sufficient to consider the movement for only one material point, which in this case coincides with the body gravity center. The method of calculating the coefficient of dynamic load on the foot (8) is described above. We introduce the coefficient K_{st}, which takes into account the total area of foot support S and the time of its contact t with the surface:

$$K_{St} = S \cdot t \tag{12}$$

The coefficient value K_{st} will be maximum at the maximum area of foot support, taking into account the period of support. The total amount of foot support reactions R_d, which create shock loads will be equal to the value of effective impact force Q_{shl} body weight. Based on this, it`s fair to say:

$$P = Q_{shl} = R_d \tag{13}$$

where $P = \Sigma P_i$ – total value of the foot load.

And from equation (12) we obtain the coefficient K_{shl} shock load for a certain phase for step cycle, which can be written as:

$$K_{shl} = K_{St} \cdot R_d \tag{14}$$

Under normal conditions, the foot is exposed to elastic deformations, i.e. it can be considered as an elastic system with one degree of freedom. Then, theoretically, if known the time t during which the pulse will be transmitted from one body to another, the impact force can be calculated by the formula:

$$Q_{shl} = \frac{p}{t} \tag{15}$$

However, in practice it`s very difficult to determine the pulse transmission time, because it depends on many different factors and can vary from microseconds to seconds. Therefore, to accurately determine the value of time, and hence the impact force is quite difficult.

To solve this problem, accept the following assumptions:

- in this research, it will consider the foot as an elastic system with one freedom degree. That is, we believe that all the deformities of the foot will be in the area of elastic, restored later.
- deformations of the considered element for design from loadings extend on all length of an element, obey Hooke's law and are proportional to deformations arising at static application of loading from the same body and in the same place.
- the proportionality of dynamic and static deformations is determined by dynamic coefficient q (6).

Given this, formula (14) will look like:

$$Q_{shl} = q \cdot \frac{m \cdot v}{t} \tag{16}$$

Taking into account expression (6) and carrying out the transformation, formula (16) is written as follows:

$$Q_{shl} = \frac{P \cdot v}{t} \tag{17}$$

where P - the total value of the load on the foot.

Given expression (13), it can be write the formula for calculating the coefficient of shock loads as:

$$K_{shl} = S \cdot P \cdot v \tag{18}$$

Researched for the coefficient of shock loads for above group of subjects. The calculation was performed by left support. Together with the calculation for this coefficient determined the length of the half-step *l*. Figure 5 shows a graph of the coefficient of foot impact loads K_{shl} on the length of the half-step *l*.

The half-step *l* calculated as the distance from the center of heel in the first phase of the step cycle to the center of heel to the same foot in the next phase, *m*. The graph shows that the maximum value of shock loads coefficient (0,584) corresponds to the maximum value of the half-step length (0,7 *m*), and therefore we can conclude that the greater for half-step length - than the greater the coefficient of foot shock loads.

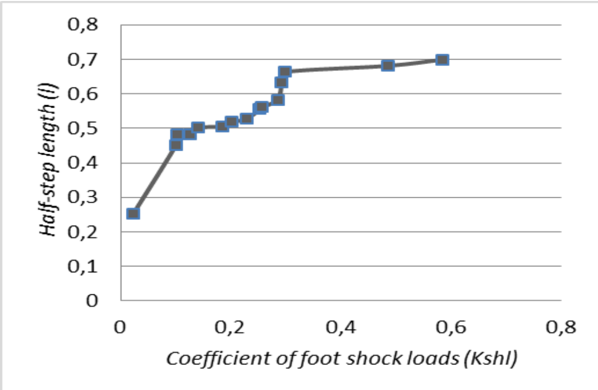

Figure 5: Graph for dependence of the coefficient of foot shock loads from the half-step length

4. Conclusions

The problem of early diagnosis for injuries and deformities the feet is relevant when choosing prevention methods, treatment, orthosis and evaluation for their effectiveness. An important design feature of the foot is its vaulted structure, which provides spring, balancing and repulsive functions. The foot, under normal conditions, is exposed to elastic deformations, i.e. it can be considered as an elastic system with one degree of freedom. It should be emphasized that the human musculoskeletal system is quite complex, but the foot as a supporting structure and part of this system is first to perceive the shock impulse of the support reaction, and its functionality directly affects the further nature for interaction with the environment. The supporting foot properties haven`t been fully researched - to increase the accuracy of their evaluation methods, it`s necessary to take into account the dynamic influence, namely short-term shock loads, on the step cycle relevant phases.

Proposed new method of analysis for stato-dynamic load, which allows to estimate dynamic and shock foot loads and is reduced to the determination for corresponding coefficients. It`s established that the maximum contact per unit time has front part (repulsion phase), then the rear part (landing phase) and the smallest - the middle foot part (rolling phase), the greater the speed and stride length, the greater the coefficient of shock loads, and their peak on the anterior and posterior departments, therefore, these areas need special attention when choosing methods of prevention, treatment and orthosis.

The practical significance for obtained results is to improve the existing methods of researching foot biomechanical parameters, by comprehensively assessing the features of standing and walking, step cycle and support properties. Considered method of researching the functional properties for lower extremities, and in the future, will be consistently improved and expand the range of its application. In particular, in order to assess the dynamic stresses arising in the feet caused by shock loads, additional researchies will be conducted using the measurement by force plate, which will increase the effectiveness of prevention, treatment and orthosis of MSS segments.

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