Introduction

Gait is the principal mode of human locomotion. Having in mind that the lower extremity and trunk joints enable only the rotary motions, the gait kinematics is a complex phenomenon. Moreover, gait as the principal mode of locomotion, poses high demands on the control system; a small surface of support, high position of the centre of gravity, uneven distribution of mass, domination of heavy segments in the upper part of the body and a high number of degrees of freedom of the entire system, are among the unfavourable features of the human locomotor system control. It ought to be also mentioned that no clear principles of the phylogenetic shaping of the organisation of joints, bones, ligaments and muscles have been found so far. The criteria of motor control involving that many degrees of freedom and redundancy of muscle drives have not been identified either. Thus, according to Winter [1], “The fact that we as humans are bipeds and locomote over the ground with one foot in contact (walking), no feet in contact (running) or both feet in contact (standing) creates a major challenge to our balance control system”.

Among the principal issues of biomechanics, scrutinized in recent years by many researchers, the effects of forces generated by individual muscle groups on the movements of specific body segments during gait are of great importance. Research in that area focuses predominantly on subjects fit psychophysically; in recent years, however, subjects with locomotor disabilities have gained marked attention in that respect.

Measuring forces at lower extremity joints during gait

Considering the human anatomy and principles of locomotor control, skeletal muscles are the only power drive of body motions. Contracting muscles produce forces acting on bone elements, to which those muscles
are attached via tendons. Determining the magnitudes and vectors of muscle forces in relation to lower extremity joints, combined with the ability of making use of them, is the key factor in gait analysis in the disabled subjects. However, in order to determine the contributions of individual muscles to the net force or net muscle torque at a given joint, the external forces acting on lower extremity joints during gait should be identified.

The solution of this problem, called in biomechanics the inverse dynamics, is now regarded as a classical method of movement modelling. Computational procedures used in the two approaches to computer simulation [2], i.e. forward dynamics and inverse dynamics, are presented in Fig. 1. Two elements in this figure are worth emphasizing: solving the inverse dynamics requires that the input should contain time functions of linear \([x(t)]\) and or angular \([\alpha(t)]\) coordinates, as well as those of the first and second derivatives; time function of external forces \([F(t)]\) represents a possible additional input to the model.

Many researchers who employ computer simulation in biomechanics consider Paul’s paper [3] as fundamental and opening new methodological perspectives in analytical biomechanics. The author adopted a four-element model consisting of articulated rigid elements. Other authors [4–10] increased the number of elements or expanded mathematical procedures.

The next step in modelling gait dynamics consisted in supplementing rigid elements with the soft tissue ones. The rigid model did not include muscles, ligaments, tendons, fat tissue, skin, interstitial fluids, etc. The viscoelastic properties of those structures imply an oversimplification of a fully rigid model.

Gruber and co-workers [11] supplemented the “ideally rigid” model with elastic elements representing the soft tissues. That model, presented in Fig. 2, has been widely accepted as the “wobbling mass model” and uses the following equations for rigid and wobbling elements:

### Rigid elements

\[
\begin{align*}
\text{Linear motion:} & \quad m_r \ddot{r}_r = \left[ \sum (F_{ik} + F_{ie} + F_{id}) + m_r g \right] \\
\text{Rotation:} & \quad I_r \ddot{\phi}_r = \left[ \sum (\Sigma F_{ik} \cdot r_{ik}) + \Sigma M_i + M_e + M_d \right] \\
\end{align*}
\]

For coupling torques:

\[
T_{wi} = \alpha \Delta \phi_i + b \Delta \phi_i
\]

\[
F_{wi} = c \text{sign}(\Delta \phi_i) \Delta \phi_i^3 + d \Delta \phi_i^3 A
\]

Surprisingly, momentary forces obtained by both models in e.g. hip joint may differ as much as 8-fold. Remembering, however, that the momentary peak force occurs in the phase of the foot–ground contact and, when the model does not include damping present in

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Figure 1. Schematic illustration of the similarities between the process of simulation and inverse dynamics analysis (redrawn from [2])
a real object, the above-mentioned difference in results is the effect of an oversimplified model.

A fair agreement of the recently reported results was due to an improved modelling of global muscle torques generated in lower extremity joints during gait [12–14]. It may thus be assumed that the curves presented in Fig. 3 adequately represent real gait-related changes in muscle torques [13].

The results of computer simulations of the inverse dynamics issues reported by the above-mentioned authors are not, however, identical despite a high similarity of global changes in lower extremity muscle torques with respect to the direction and tendencies of those changes. The differences in those results may be due to uneven reliability of data input into the model. The variables used to construct models may be classified as follows:

Figure 2. The three-linked model with wobbling mass [11]

Figure 3. Normalized joint moments (N m/kg) about ankle, knee and hip in frontal, sagittal and transverse planes under global and local coordinates. X-axis represents relative time, with 0% indicating heel contact and 100% indicating toe-off. The curves represent averaged global joint moments (—) and averaged local joint moments [13]
1. Reliable and easily attainable,
2. Reliable but attainable only by employing sophisticated methods (e.g. ground force components),
3. Unattainable intravitaly,
4. Non-measurable, due to the fact of being insufficiently defined.

The values of variables from the last two categories are usually obtained post-mortem or assessed by the trial-and-error approach. The forces generated by individual muscles during gait are necessary in modelling but impossible, so far, to be measured directly.

Determining forces generated during gait by individual muscles of lower extremities

Static optimisation is among the most popular methods of determining forces generated by individual muscles in diverse motor activities. The first stage of this method consists in solving the problem of inverse dynamics for discrete values of time, usually equal to the frequency of data recording. The next step is the partition of global muscle torques into individual muscles at consecutive moments of time by employing procedures leading to extreme values of appropriately defined object functions of the process under study. A comprehensive overview of the most popular object functions and of numerical procedures suitable for static optimisation was presented by Tsirakos et al. [15]. A high numerical stability of algorithms, simple course and short computational time, especially for two-dimensional models, are the prime advantages of static optimisation. Computing the time courses of forces of 8 muscles in a planar three-component model of lower extremity in a single gait cycle took only 5.3 s [16], and force characteristics of 9 muscles executing leg swing took 7.2 s each [17]. Stable time courses of all muscle forces were obtained in complex three-dimensional models consisting of as many as 47 or 43 muscles [18, 19].

The principal drawback of static optimisation is the presumed optimal exertion of muscles at every moment of a given motor task; that presumption may greatly affect the results and is difficult to be experimentally verified. In effect, the obtained discrete characteristics of muscle force may not correspond to its real course. Yet, static optimisation was demonstrated to be a valid and reliable tool in studying gait [20].

Contributions of individual muscles to the global force of various muscle groups may also be determined from EMG signals [7, 21, 22] but the respective biomechanical models of gait are not popular due to serious errors associated with force assessment in that way, especially in the phase of signal recording. Namely, superficial EMG does no enable recording signals from all individual muscles involved in the motor act; moreover, active muscle fibres located under the muscle examined distort the specific signal.

Possible use in implanted electrodes poses great difficulties and as a routine practice would be rather unreasonable as the access to all muscles engaged may not appear feasible. The lack of experimental data which might reveal the relations between EMG signal variables and changes in muscle force under dynamic conditions, i.e. muscle-driven movements of lower extremities, is another drawback in the application of EMG to determining forces generated by muscles during gait. On the other hand, changes in EMG amplitudes were linearly correlated with muscle force in isometric contraction (static conditions) in the range of 40 to 80% MVC.

EMG records are thus mostly used to verify the reliability of numerical gait simulations [20, 22–24]. The latter authors designed a human musculoskeletal model by using SIMM (Software for Interactive Musculoskeletal Modeling; MusculoGraphics, Inc., USA) package and demonstrated that multijoint muscles – knee and hip extensors – are essential in maintaining body stability in the initial, single support phase of gait. Moreover, the re-distribution of muscle power in the entire gait cycle enabled trunk and lower limb motions. The intuitive obviousness of results confirms the usefulness of the software and of the numerical procedures employed.

Nevertheless, many authors continue the analytical approach to assess the contributions of individual muscles to the global strength. The approaches include the input of measurable variables like: (1) global muscle torque of a single muscle group under static conditions; (2) muscle lengths measured anthropometrically; (3) muscle cross-sections determined ultrasonographically; (4) moments of inertia of segments by applying forced vibration, and (5) experimental data of rigidity and damping. However, the input of the following variables into the model remains unsolved: (1) stresses transmitted by individual muscles; (2) rigidity of the anatomical muscle–ligament interface; (3) activation, i.e. the function of muscle stimulation, and (4) criteria of optimum control of muscle work and of the whole body movements. An advanced simulation procedure was present-
ed by Anderson and Pandy [25] or Pandy and Zając [23]. The authors started with simulation and performed analytically a dynamic optimisation solution. Next, they collected some data recorded in corpses, in several subjects, as well as data from earlier publications on strength–speed relationships in selected muscle groups, they simulated the contributions of individual muscles to the ground reaction forces and accelerations of individual body segments. The results, which seemed to be close to the real ones, enabled analysing the contributions and significance of those segments in maintaining equilibrium in the support phase and in forward driving the body during gait.

Both papers mentioned above demonstrated the way to compute forces of individual muscles by applying forward dynamic simulation, but employing different numerical approaches. The first approach, a continuation of their earlier report [26], was based formally on the optimal control theory (Pontryagin’s principle); the authors maximised the Hamiltonian function and solved the resulting boundary problem by integrating status variables and conjugated variables in opposite directions. By assuming a linear relationship between muscle activation and its stimulation, a bang-bang control, represented by stimulation, was obtained. Despite the physiological correctness of that solution, the method has been rarely used in biomechanics due to unstable integration of conjugated variables [27], difficulties in physical interpretation of variables and a likely appearance of singular control.

The other approach, first described by Pandy, Anderson et al. [28], avoided the need of solving the boundary problem by transforming the optimal control into parameter optimising problem. The control, expressed also by muscle stimulation, was spread over a net of nodal points, uniformly distributed over time, and forming a discrete set of independent variables subject to optimisation. In that way, the issue was reduced to seeking nodal control values which minimise the integral quality index defined for a given motor task and, at the same time, maintain the constraints forced by the state equations containing the co-ordinates and generalised velocities, muscle forces and muscle activation levels as dependent on their stimulation. These equations were integrated conventionally by using the Runge-Kutta algorithm and the optimal values of control variables were obtained by sequential quadratic programming (SQP). The common use of parametric optimisation greatly reduces the computation time which is a derivative of integrating state equations at every step of the iterative SQP. The computations of two-dimensional models by using a PC take up to several days [29], and up to several months in the case of complex three-dimensional models [24].

A hybrid solution, which combines the features of direct and inverse dynamics, was presented by Menegaldo, Fleury et al. [30]. The first step was the inverse dynamics solution followed by defining a function of errors between the computed global muscle torques at given joints and their optimised equivalents resulting from muscle forces. That function was attached to the integral quality index of the process. Optimisation was conducted by SQP, maintaining the differential constraints describing the contraction and activation dynamics. A substantial reduction of computation time was achieved by eliminating the co-ordinates and generalised velocities from the state vector. Inasmuch highly promising, the arbitrary weights attributed to the

Figure 4. Contributions of inertial forces (Inertial), of centripetal forces (Centrifugal), of the resistance to gravity provided by the bones and joints of the skeleton (Gravity), and of muscle and ligament forces (Muscles + Ligaments) to support during normal gait. The thick gray line is the vertical ground-reaction force obtained from the dynamic optimisation solution for gait. Total (thin black line) was obtained by summing Inertia, Centrifugal, Gravity and Muscles + Ligaments. The mean vertical ground-reaction measured for the subjects are shown as a dotted line (Subject Mean). The marked kinematic events are: HS – heel-strike, FF – foot-flat, CTO – contralateral toe-off, HO – heel-off, CHS – contralateral heel-strike, MO – metatarsal-off, and TO – toe-off. BW is body weight (BW = 696 N for the model and the subject mean) [25]
two components of the quality index represent some weakness of the approach.

Ackermann and Schiehlen [16] presented a novel procedure of determining forces for individual muscles based on the inverse dynamics solution and called it the ‘extended inverse dynamics’. They modified the classical procedure by including force dynamics and muscle activation, as well as an integral quality index for the entire motor task. Parametric values of muscle forces were taken as optimised variables and an algorithm of sequential programming was used in optimisation. Since the quality index was related to the activation and stimulation of the muscles, time functions of these variables constituted an integral part of computations. The first function was obtained by solving the force–length–velocity relationship for the contractile elements of the muscle preceded by numerical differentiation of muscle force by time, and the other one resulted from differential relationship between activation and stimulation preceded by differentiation by activation time. That method, free of the drawbacks of the classical inverse dynamics, proved more efficient numerically compared with parametric optimisation based on simple simulation. The time needed to generate functions of muscle forces over a single gait cycle for a planar model of lower extremity including 8 muscles amounted to about 12 h [16]. Solving an identical task for leg swing with the use of a planar model which included 9 muscles took over 2 h [31] but in this case a much simpler quality index was used.

A highly advanced software for analysing muscle force (SIMM; Software for Interactive Musculoskeletal Modelling) was designed by Loan, Delp, Smith, Blake and Megelan (MusculoGraphisc Inc., USA) with a significant contribution of van der Helm. The model utilised data from numerous publications and probably also the author’s own data. Gait simulation based on ground force values was conducted by the Laboratory for Functional Anatomy, Biomechanics and Motor Control in Copenhagen (Denmark). An example graph of knee joint muscle torques as functions of knee joint angle during gait is presented in Fig. 5. Inasmuch some doubts exist as to whether the author did not manually adjust the force curves as functions of the joint angle so as to obtain a satisfactory concordance with the resultant global muscle force at the hip, knee and ankle joints, the software and gait model, as well as other products offered by the MusculoGraphics, enable diverse simulations, e.g. of changes in muscle length, stress in bones, etc.

![Figure 5. Muscles torque changes of the selected knee extensors during normal walking. Simulation was provided by SIMM](image)

**Clinical examples of the effects of force reduction of selected muscle groups on the gait mechanics elements**

Bennett, Ogonda et al. [32] analysed the effects of two kinds of hip joint surgery – minimally invasive and conventional, by employing variables of gait kinematics, i.e. downward pelvis displacement during gait; adduction/abduction, flexion/extension and rotation at the hip joint; flexion/extension at the knee joint; foot adduction/abduction; flexion/extension at the ankle joint. Gait velocity, step length, double step length and the duration of the foot–ground contact in the support phase did not differ significantly in both groups on the 2nd day or 6 weeks post-surgery, while rotation at the hip joint differed significantly between groups as well as from the normal values.

Similar values of gait velocity, step length and support duration were noted in women suffering from myofibrosis [33] and ground reaction forces in those women and the control ones were alike. The authors concluded that the pattern of recruitment of lower extremity muscles in myofibrotic women differs from that in healthy subjects; specifically, plantar flexion in myofibrotic women requires a markedly greater muscle work than in the healthy ones, which makes walking very fatiguing.

A similar study was conducted in children with cerebral palsy and, thus, crouch gait [34]. That kind of gait is due to abnormally short hamstring or the semimembranosus muscle spasm. The authors recorded changes in muscle length and simulated its shortening rate; they
demonstrated that those children subjected to hamstring elongation who made no use of elongation and, thus, of generating higher forces by that muscle and did not improve the gait-induced shortening rate, were incapable of improving their gait quality. The authors identified the criteria of selecting children who may significantly improve their gait post-surgery. In another study, Goldberg, Öunpuu et al. [35] studied gait in children with cerebral palsy, aged 10 ± 2.8 years, subjected to transplantation and elongation of the rectus femoris muscle tendon, elongation of the gracilis and gastrocnemius muscles, de-rotational osteotomy and corrective foot surgery. The analysis of gait dynamics revealed an improvement in the vertical component of ground reaction force and in the motion range in leg swing. The authors showed that anatomical correction of the rectus femoris muscle significantly improved the single support phase, stabilised the gait and made it more efficient. Moreover, they showed that the excessive knee contracture in the support phase was due not only to a functional insufficiency of the rectus femoris muscle but also to a contracture and low strength of the gastrocnemius muscle. Since not all the children exhibited a post-surgical improvement, the question as to whether the transfer of the rectus femoris muscle is the only factor that might improve the stiff knee in children with cerebral palsy, remained unanswered.

Wu, Su et al. [36] presented a mathematically more advanced approach to gait analysis following ankle arthrodesis. A three-component foot model was assumed and 11 markers were used to delimit the components. A kinematographic recording was conducted together with recording ground reaction forces and EMG signals from 5 lower extremity muscles. As compared with healthy subjects, patients with ankle arthrodesis had significantly smaller motion range of the distal part of the foot, greater mobility of the frontal part of the foot in the sagittal and transversal planes, a generally greater stiffness of the distal part of the foot combined with a greater engagement of foot muscles and a decreased EMG activity of large muscle groups, e.g. rectus femoris or soleus muscles.

Gait kinetics and kinematics was also reported for other diseases, e.g. parkinsonism [37]. Inasmuch the advancement of the disease determines the magnitudes of changes, the general characteristics of gait mechanics are disease-specific. Similar findings were reported for subjects with a shorter [38] or amputated [39] lower limb.

Determining forces generated during gait by individual muscles of lower extremities by using artificial neural network modelling and simulation

Artificial neural network (ANN) is an artificial intelligence method used in mathematical modelling and its applications in diverse areas, especially in biology and medicine, are steadily progressing. The achievements and possibilities of ANN in biomechanics were presented e.g. by Simon [40], who discussed the status of studies on gait biomechanics and pointed to advantages of gait analysis for clinical purposes, as well as to persisting limitations of ANN. Simon’s conclusions confirm and support an earlier review on ANN applications to gait analysis [41] and seem reasonable in view of recent reports [42, 43]. Lees [44] reviewed world tendencies in applications of biomechanics to the analysis of movements in sports and emphasised the role of ANN. Worth mentioning are also applications of ANN to the functional electrostimulation supporting the gait in patients with parapareses [45, 46].

Apart from the use of ANN in general gait analysis, like gait velocity or step frequency [47], distribution of forces generated by foot pressure on the ground [48] or ground reactive force [49], in the available literature there were found six reports pertaining to the use of ANN in computer simulation and modelling muscle work during gait. Specifically, the authors discussed the modelling of muscle activity as a basis of modelling changes in kinematic gait parameters [50, 51], modelling muscle work of lower extremities by using kinematic and kinetic parameters [52, 53], or assessing muscle strength from EMG signals [13, 54].

Sepulveda, Wells et al. [50] presented a different approach to the construction of models; they recorded EMG potentials from 16 muscles during a walk at natural velocity together with muscle torques and changes in angles at the hip, knee and ankle joints. They designed a network which represented relationships, although not quite accurately, between the changes in activities of muscles and the movements of lower extremities. They observed the effects of reducing the stimulation of the soleus muscle by 30% and of the total exclusion of the rectus femoris muscle. Best results were obtained by relating in ANN the EMG signals with muscle torques.

Tucker and White [51] studied changes in the tonus of five muscles in relation to the gait velocity and step frequency. Muscle activity was determined by EMG and kinematic variables on a mechanical treadmill. Hel-
A more advanced network was presented by Prentice, Patla et al. [53]. They fed the following data: step frequency, lengths of individual steps, durations of support and of extremity swings. Twelve gait forms were studied on a track equipped with obstacles forcing the subjects to walk with varying velocity, step length and knee lifting height. All those variables were input into the network, while EMG variables of eight muscles (gastrocnemius medialis, soleus, tibialis anterior, peroneus longus, biceps femoris, rectus femoris, gluteus medius, erector spini) during gait constituted the output. The simulated activities of individual muscles were highly satisfactory and the network was capable of reproducing the real activity patterns of all eight muscles throughout the double step cycle. Inasmuch momentary EMG values differed up to 25% in 25 out of 96 combinations of muscles and gait forms, it could be concluded that ANN proved a reliable tool in the modelling and simulation of muscle work based on principal variables of gait kinematics.

EMG signals were also used to assess the stress forces in the tendons. Savelberg, Herzog [54] designed a multilayer network, fed by EMG signals from the gastrocnemius muscle, which generated force values measured by a gauge fixed to the tendon. The study, conducted on three cats, included three experiments:

- Simultaneous recording of force and EMG at constant gait velocity on a mechanical treadmill,
- EMG was recorded, force was measured and compared with that predicted by ANN simulation (the treadmill ran at three different velocities),
- Forces in one animal were predicted from EMG values recorded in the other two cats running at various velocities.

This is, so far, the only report found in the literature in which muscle forces predicted from EMG signals during gait by ANN simulation were verified experimentally. The results may be regarded as reliable as the simple and multiple correlations between the predicted and observed values were high in all versions and ranged from $R^2 = 0.71$ to $R^2 = 0.98$.

Liu and Lockhart [13] attempted at creating a network capable of reproducing muscle forces during gait from EMG signals recorded in working muscles. The bioelectric activity and force of the soleus muscle were recorded, the experimental protocol from the previous experiment being exactly reproduced. The objective of the study was to make use of the experience in the construction of ANN and to apply advanced mathematical procedures to identical experimental conditions. The authors emphasized their achievement consisting in constructing a network with three latent layers with a single input containing multiple EMG(t) signal cells and a single F(t) output, capable of reproducing muscle forces with an accuracy equal to $R^2 = 0.9$ and RSE $< 15\%$ by processing data from various animals.

Conclusions

Summing up, the following issues remain to be solved:

- Models of distributions of muscle forces in diverse muscular and neural pathologies;
- The SIMM software and the available model of lower extremity movements in gait are not suited for reproducing pathological gait resulting from a deficient muscle strength or neuromuscular co-ordination, as the software was designed for normal gait;
- It is not possible to design a reverse simulation, i.e. predicting muscle activity from the mechanics of pathological gait.

The available literature offers several mathematical procedures, including ANN, for modelling muscle work in specific movements of the whole body or of its segments, the ANN being particularly useful in studying biological objects for the following reasons:

- It enables analysis of non-linear relationships between parameters and between subjects;
- ANN procedures are distribution-free;
- Both continuous and categorised variables can be easily processed;
- Outliers and missing data are easily managed;
- ANN procedures are noise- and interference-resistant.

The principal advantage of ANN is the capacity to conduct simulations and designing models having a limited number of observations and a large number of variables at the disposal. In contrast to deterministic methods, ANN techniques do not require complicated procedures of establishing physical modelling followed
by mathematical description, which makes ANN a handy analytical tool in practically any experimental area. Thus, ANN gains increasing application in biomechanics and shall be employed in future research.

References


HUMAN MOVEMENT
A. Wit, A. Czaplicki, Inverse dynamics and ANN


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