Influence of modified muscle morphology and activity pattern on the results of musculoskeletal system modelling in cerebral palsy patient

JUSTYNA OGRODNIK*, SZCZEPAN PISZCZATOWSKI

Department of Biocybernetics and Biomedical Engineering, Faculty of Mechanical Engineering,
Białystok University of Technology, Białystok, Poland.

Purpose: The aim of the present study was to evaluate the influence of modified morphological parameters of the muscle model and excitation pattern on the results of musculoskeletal system numerical simulation in a cerebral palsy patient. *Methods*: The modelling of the musculoskeletal system was performed in the AnyBody Modelling System. The standard model (MoCap) was subjected to modifications consisting of changes in morphological parameters and excitation patterns of selected muscles. The research was conducted with the use of data of a 14-year-old cerebral palsy patient. *Results*: A reduction of morphological parameters (variant MI) caused a decrease in the value of active force generated by the muscle with changed geometry, and as a consequence the changes in active force generated by other muscles. A simulation of the abnormal excitation pattern (variant MII) resulted in the muscle's additional activity during its lengthening. The simultaneous modification of the muscle morphology and excitation pattern (variant MIII) points to the interdependence of both types of muscle model changes. A significant increase in the value of the reaction force in the hip joint was observed as a consequence of modification of the hip abductor activity. *Conclusions*: The morphological parameters and the excitation pattern of modelled muscles have a significant influence on the results of numerical simulation of the musculoskeletal system functioning.

Key words: *modelling, muscle, cerebral palsy, joint reaction force*

1. Introduction

Cerebral palsy (CP) is a group of symptoms associated with a variety of etiology of the central nervous system damage occurring in the developing fetal or infant brain. Due to the frequent occurrence of the disease (2–2.5 per 1000 live births), it is a significant diagnostic, prognostic, therapeutic and social problem [21]. The incorrect function of the central nervous system leads to secondary musculoskeletal problems [3]. The most common consequence of cerebral palsy is the spasticity which represents nearly 85 percent of all CP cases [22]. Several definitions of spasticity appear in the neurological literature, however, the most commonly used is probably the proposal presented by Lance in 1980 which says that it is "a motor disorder, characterised by a velocity-dependent increase in tonic stretch reflexes (muscle tone) with exaggerated tendon jerks, resulting from hyperexcitability of the stretch reflex as one component of the upper motor neurone syndrome" [16]. Spastic muscles in CP children are characterized by different morphology and excitation patterns compared with the muscles of healthy peers. The most evaluated and analysed parameters of spastic muscles include: muscle length, fascicle length, cross-sectional area, fascicle angle, thickness and lower resting sarcomer length, muscle belly volume [11], [18], [20], [24]. The reduced size of muscles of lower limbs was estimated as a result of research conducted with the use of MRI and ultrasound scans [18], [20], [24]. A review by Barrett and Lichtwark concerning the morphology of muscles showed a reduction of the muscle volume of

about 20–57% in the limbs of CP children compared to healthy peers [2]. Shortland et al. [24] reported that a reduction of the muscle volume was observed in the whole lower limb, however, the muscles in distal parts were most affected. According to the research by Lampe [15], the total muscle atrophy was more visible in the lower leg (28%) than in the upper leg (16%). It is commonly believed that cerebral palsy occurs heterogeneously, depending on the type of brain damage [11]. The results of Handsfield et al. show that the muscles in CP patients are small, however, overall the magnitude of muscle deficit volume, length and cross-section area vary depending on the severity of the disease. The most affected muscles in CP patients included the soleus, gastrocnemius, tibialis anterior, semimembranosus, all plantarflexors, dorsiflexors and hamstrings [11]. The results of the dynamic analysis conducted using isokinetic dynamometers or the evaluation of gait show additional differences in the functioning of the spastic muscle. An analysis using the motion capture data, EMG and SIMM software in the research by Krogt et al. revealed a decreased velocity during the stretch and an increased muscle activity in the swing phase. Spastic muscles were about 30% slower and their activity was 3 times higher in the swing phase compared with the muscles in typically developing children [14].

The biomechanical analysis of pathological gait in CP [7], [8], [25] is a challenge because of the complexity of consequences of the disease for the locomotor system. Therefore, modelling and numerical simulation are increasingly used for the advanced pathological musculoskeletal system analysis. The musculoskeletal system modelling delivers new information about both normal and pathological gait excessive internal hip rotation [1], stiff knee gait [13], surgery consequences for the muscle-tendon lengthening or validation of treatment strategies [5]. Most biomechanical studies use standard models based on muscle architecture, musculoskeletal geometry and neuromuscular excitation patterns from unimpaired subjects. Developing patient-specific models based on the results from magnetic resonance images or ultrasound scans allows muscle lengths or moment arms to be realistycally estimated in a range of body positions during movement. However, this method has its disadvantages because it requires extensive imaging protocols to capture the muscle and joint geometry in many limb configurations [23]. Hainisch et al. [10] created a musculoskeletal model for CP patients by combining MRI data (for selected segments and muscles) with motion capture data. This model was examined using OpenSim software, which allowed the

muscle force to be calculated for selected muscles. The use of the MRI data and the OpenSim system made it possible to better scale the model of skeletal muscle by changing the position of the muscle attachment. The results indicate that developing subject-specific models is useful, however, the methodology should take into account the information of abnormalities of the musculoskeletal system such as the muscle spasticity and incorrect activity patterns.

A review of literature indicates that numerical simulations of the musculoskeletal system may deliver clinically useful information about the pathology gait. However, in the case of spastic muscles, it is necessary to improve the methodology of modelling by taking into account morphological changes and reconstructing changed rheological properties and abnormal activity patterns (excitation patterns).

The aim of the present study was to estimate the influence of modified muscle morphology and excitation pattern on the results of musculoskeletal system modelling in cerebral palsy patients.

2. Materials and methods

The concept of this study is based on the use of a model of the musculoskeletal system to evaluate selected biomechanical parameters (muscle force, joint reaction force) during the pathological gait of a CP patient. The results obtained with the use of a standard model of the musculoskeletal system available in the repository of the AnyBody system (STEP 1) will be compared with the results of a simulation conducted on a modified model with changed morphological parameters and/or activity patterns for selected muscles (STEP 2).

Patient's data

The model presented in this paper was developed based on the results of a study on a 14-year-old young female patient (body weight 39 kg; height 152 cm) suffering from cerebral palsy (diplegia) obtained with the use of the motion capture system (Vicon). The motion camera system was integrated with two force platforms of Kistler measuring the ground reaction forces and the surface EMG system used for the electrical muscle activity registration.

The study was conducted for diagnostic needs of the patient, so in addition to the source data also a clinical report containing the patient's anthropometric data, results of kinematic analysis and the information on the activity of the muscles (rectus femoris, biceps femoris, tibialis anterior, gastrocnemius, gluteus maxi-

mus, hamstring, soleus muscles) were available. The results of the clinical report showed spastic diplegia and consequently an increased pelvis anteversion, severe knee valgus, excessive flexion of knee and hip joints along with their internal rotation during the whole gait cycle, and a strong adduction in the hip joints in the stance phase.

Modelling

For the analysis of the musculoskeletal system the AnyBody Modeling System was used (AnyBody Technology A/S, Aalborg, Denmark). The MoCap model used for the gait analysis consisted of 17 body segments altogether (7 odd: pelvis, sacrum and 5 lumbar vertebras, and 5 even: right and left femur, patella, tibia, talus, foot) joined by 11 joints (L5-S1 and left and right hip, knee, patella/femur, talocrural and subtalar) with 21 degrees of freedom. All segments were modelled as rigid bodies. The model of each lower limb contained 56 muscles (muscle-tendon) divided into 159 branches. The Hill's model applied in the AnyBody system to describe the muscle consisted of a contractile element (CE – active force generated by the muscle) and elastic elements arranged in parallel (PE – stiffness of muscle fibre) and in series (SE – tendon stiffness). Force characteristics (force –velocity and force–length relationships) of muscle-tendon unit depend on both its architecture and its intrinsic properties. Hill's model is sensitive to parameters, especially fibre length, physiologic cross-sectional area and tendon length [12]. This way Hill's model of muscle-tendon unit can be adjusted for the needs of normal as well as pathological muscle functioning. Van der Krogt et al. have used such a model for neuro-muscular simulation of contracture and spasticity assessment in children with cerebral palsy [26]. To adapt muscle model to individual characteristics of pathological muscles the following parameters can be used: muscle belly volume, pennation angle, muscle fibre length, physiological cross-sectional area (PCSA), tendon length and some other (e.g., peak isometric force) [12].

STEP 1 – preparation of initial model

The standard model (M0) was a basis for further research. The data of the current position of markers and the ground reaction force at selected moments of the gait cycle were downloaded from an input file in the C3D format containing the patient locomotion test results. The rigid segments were scaled using the patient's anthropomorphic data (height, weight, width of pelvis, length of thigh, shank and foot). The muscle

attachment location and its morphological parameters were scaled with the whole model (Fig. 1).
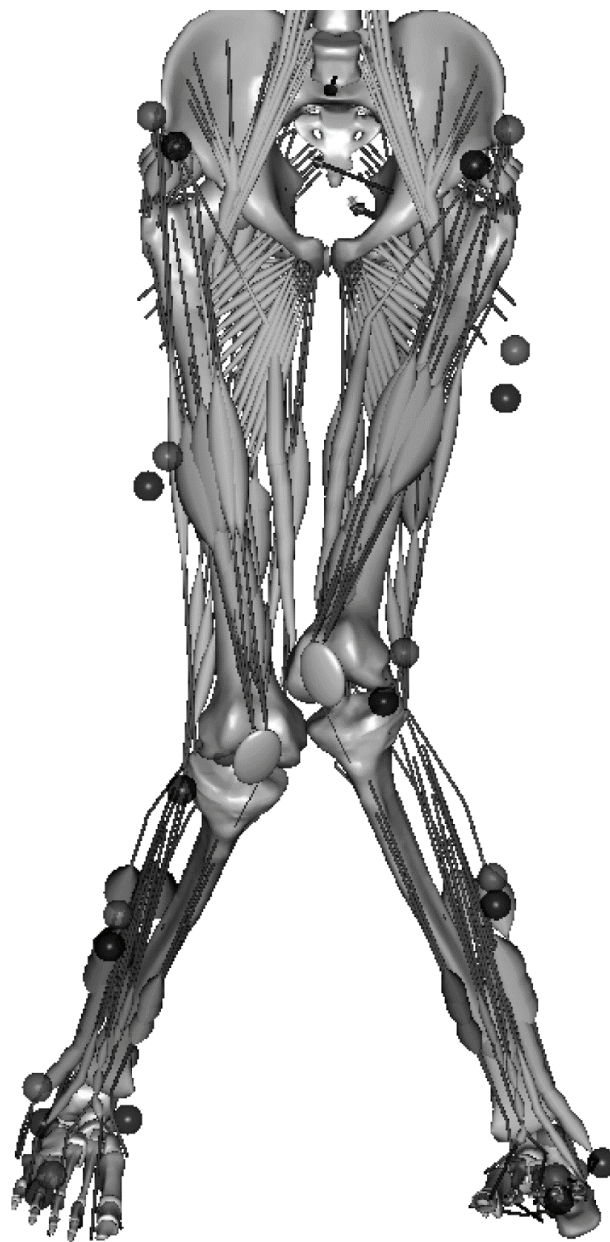

Fig. 1. Visualization of the scaled musculoskeletal model for a 14 year-old young female patient in the AnyBody system

STEP 2 – modifications to the model

The standard model (M0) was modified in a few ways. The purpose of the first variant (MI) was to change the morphological parameters of selected muscles. In the second variant (MII), the pattern of muscle excitation (activity) was modified to reconstruct some features of the spastic muscle [16]. The modification to the model in variant MIII consisted in a combined change in the muscle morphological parameters and excitation patterns.

Variant MI – change in muscle parameters

The morphological parameters of the muscle model were modified on the basis of the results of research on spastic muscles available in the literature. When selecting muscles to be modified the Handsfield study [11] which indicated the muscles that are subject to the largest pathological changes in particular types of cerebral palsy was applied. The most changed muscles in the case of diplegia and crouch gait are presented in Table 1 and were modified in model MI. In the case of selected actons the length of fibres of the contractile part was decreased (with unchanged length of the whole muscle) and the muscle volume in relation to the parameters of the standard model was reduced, while the range of changes (Table 1) was determined on the basis of averaged values of deficits presented by Handsfield [11], Lampe [15], Mohagheghi [20] and Shortland [24].

Variant MII – change in muscle activity

It is assumed that the spastic muscle is active during elongation [16], and its minimum excitation is associated with the velocity of muscle elongation, according to the relationship:

$$A_{\min} = \min\{1; b \cdot \dot{L}_m\} \qquad (1)$$

$A_{\min}$ – minimum muscle excitation,
b – proportionality coefficient, $b = 2$ was arbitrarily assumed,
$\dot{L}_m$ – velocity of muscle elongation.

The changes in the muscle excitation pattern described were applied in the case of a selected muscle group:

Variant MIIA – rectus femoris, biceps femoris, gastrocnemius, tibialis anterior – the selection was made based on the EMG test of evaluated patient;

Variant MIIB – as in variant MIIA plus adductor muscles of the hip. Taking into account the fact that kinematic parameters of the evaluated patient's gait indicated an excessive adduction of lower limbs, the muscle group selected on the basis of the EMG test findings was complemented by thigh adductors which were not tested by the EMG. An incorrect activity was applied in the case of all adductors (variant MIIB) or

Table. 1. Characteristics of the changes in morphological parameters of muscle model.
Percentage value of decrease in particular parameters of model MI in comparison to standard model M0

Muscle	GA*	SO	TI	RF	ST	SM	BF	AM	AB	GMa
Fascicle length	12%	8%	6%	14%	19%	7%	3%	6%	2%	2%
Muscle volume	36%	38%	47%	33%	44%	33%	30%	21%	19%	11%

* GA – gastrocnemius, SO – soleus, TI – tibialis anterior, RF – rectus femoris, ST – semitendinosus, SM – semimembranosus, BF – biceps femoris, AM – adductor magnus, AB – adductor brevis, GMa – gluteus maximus.

Table 2. Variants of musculoskeletal model

M0 Standard model			
MI Modified morphological parameters of muscles			
MII Modified muscle activity			
MIIA Choice of muscles based on the electromyography (EMG) test: TI, RF, BF, GA	MIIB Changed activity of muscles tested using the EMG (TI, RF, BF, GA) with an additional abnormal activity of three hip adductors		
	MIIB_b only AB	MIIB_l only AL	MIIB_m only AM	MIIB_{b+l} AB and AL
MIII $\sum[(MI + MII)]$			
MIIIA Modified morphological parameters and activity pattern of muscles: selection of muscles based on the EMG test (MI+MIIA)	MIIIB Modified morphological parameters and activity pattern of muscles: selection of muscles based on the EMG test plus three hip adductors (MI+MIIB)		

only one of the adductors: MIIB$_b$ – only adductor brevis, MIIB$_l$ – only adductor longus, MIIB$_m$ – only adductor magnus. Due to the lack of unambiguous information on the activity of the medial group of thigh muscles and the fact that adductor magnus in the stance phase can work mainly as a hip joint extensor [16], an additional variant of the modification was made where the activity was changed only for adductor longus and adductor brevis with the omission of adductor magnus (MIIB$_{b+l}$ variant).

Variant MIII – changes in morphological parameters and activity of selected muscles

The last variant of model modification consisted in combined changes in morphological parameters of the previously described muscle group and the excitation pattern of four muscles tested using the EMG (variant MIIIA), and additionally three hip adductors (variant MIIIB).

The list of all model variants is presented in Table 2.

Analysis of the model

The models thus prepared and scaled constituted the basis for dynamic analyses. Experimentally recorded marker trajectories and GRFs patterns were imported into the AnyBody. The joint moments were calculated using the inverse dynamics analysis and thereafter, a muscle force distribution problem was solved using optimization procedure with the 3rd order polynomial muscle recruitment criterion [27]. EMG data available for six muscles in the clinical report were not included in dynamic simulation directly. However, such data can be used for evaluation of the muscle activity period obtained as a result of numerical analysis.

The hip joint was modelled as a frictionless, ball-in-socket 3-DOF joint, where the femoral head is treated as a ball and the acetabulum as the socket (spherical joint). The joint reaction was calculated as a result of the forces of gravity and inertia as well as muscle forces and external loadings acting on the bone element forming the joint. Finally, the resultant force was calculated in the femur coordinate system and normalized to subjects' body weight (BW).

Considering the parameters describing the muscle functions (e.g., change in muscle elongation, muscle excitation, muscle force), the information about the activity of the muscle whose model was modified and the influence of such muscle on the entire musculoskeletal system was obtained. Finally, the influence of particular model modifications on the active muscle force and reaction in the hip joint during the gait cycle

were analysed. The above-mentioned parameters constituted the basis for assessment of the influence of the modified muscle characteristics on the functioning of the cerebral palsy patient's musculoskeletal system model.

3. Results

3.1. Active muscle force analysis

In order to assess the influence of the modified morphological parameters and excitation of selected muscles on the functioning of the musculoskeletal system model, the active muscle force patterns generated in particular actons were evaluated.

Effect of modification of morphological parameters (variant MI)

The results for the muscles in which the morphological parameters of the muscle model was changed (Variant MI) were presented in Fig. 2a. Analogous results for the selected muscles whose model was not modified were presented in Fig. 2b.

In all ten actons* in which morphological parameters of a model were modified, differences in the active force patterns were observed. Although there was a slight variation, the active force generated by particular muscles generally decreased. The range of change was between 10 and 50% of values obtained for a standard model M0. It can also be seen that the active force of biceps femoris caput longum (BFl) muscle increased in the swing phase (75–100% of the gait cycle), however, it was not noticeable in other muscles with modified morphological parameters.

An exception to the abovementioned observations is the behaviour of gastrocnemius (GA), adductor magnus (AM) and gluteus maximus (GMa). In the case of GA, despite the reduction of the muscle volume by 36%, the active force increased in the stance phase (10–35% of the gait cycle) in comparison with the standard model. When we compare this result with a reduction of the force in SO occurring at the same time, it may be concluded that there was compensation within the triceps surae. When two analysed back muscles of the thigh acting on both hip and knee joints (BFl, ST, SM) were weakened during the whole stance phase, the active force developed by the

---

* Acton – Muscles or muscle parts with point-to-point attachments that generate moments about a single joint axis.
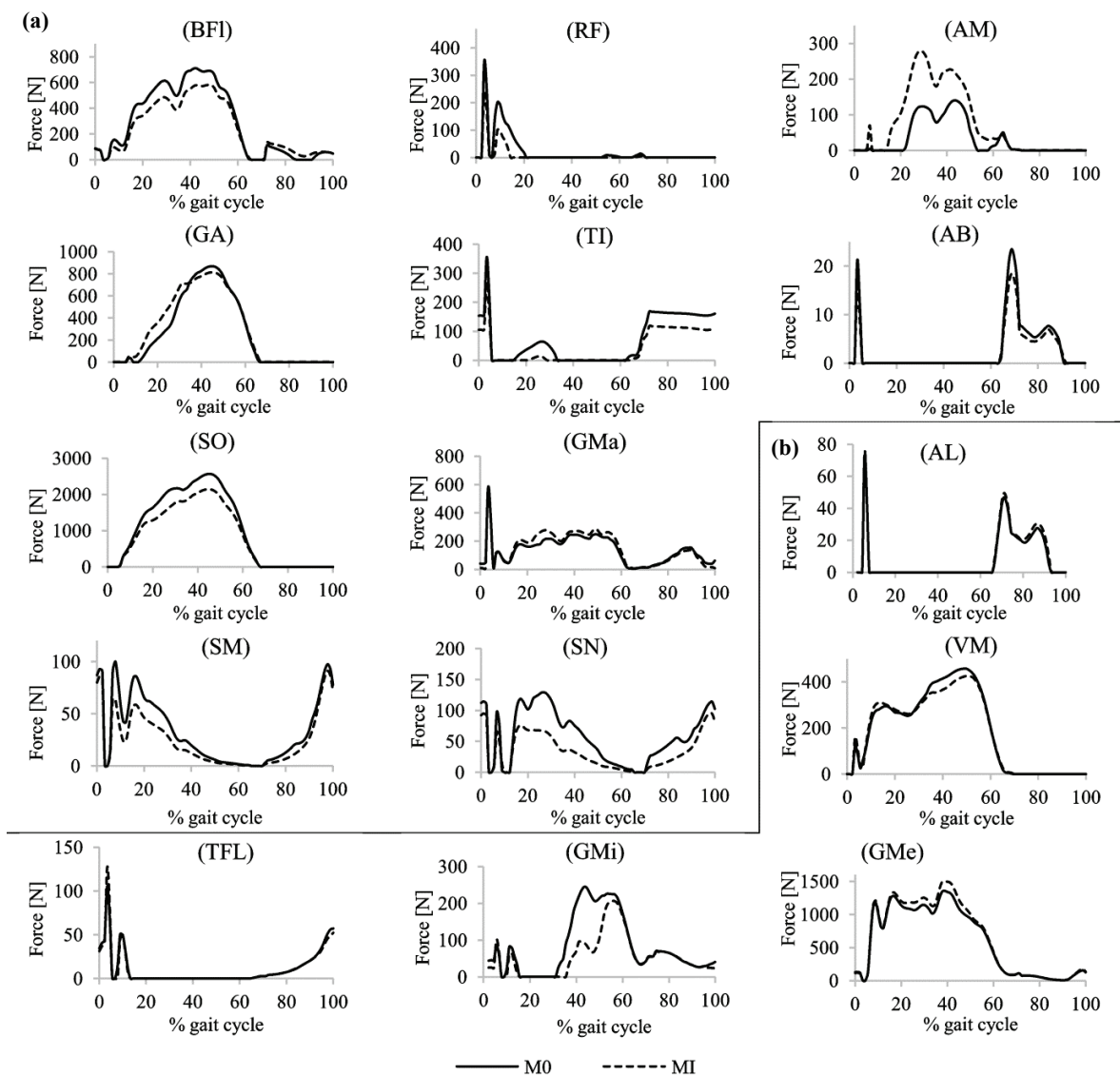
Fig. 2. Muscle active force waveforms for the muscles in which morphological parameters were modified (variant MI) (a) and without any changes in their models (b). The results obtained for the first variant (MI) of a model modification are presented with references to the results obtained for the standard model (M0). Abbreviations of the muscles as defined in the text

strongest hip extensor (GMa) remained initially unchanged (initial double stance – 0–19%) despite the reduction of its morphological parameters (volume reduced by about 11%) and increased by 12%–13% in the further part of the stance phase (20–69%). In addition, the active force generated by the AM, being also a very strong extensor of the hip joint, increased during the whole stance phase. It shows that these muscles took over a part of the tasks of other hip extensors doing the work necessary to maintain the upright posture.

A change in morphological parameters of the selected group of 10 actons forced the change of active force values generated in other muscles, including those whose model was not modified (Fig. 2b). The active force increased in the vastus medialis (VM)

during initial double stance phase, in the gluteus medius (GMe) during the stance phase and in the adductor longus (AL) during swing phase. However, the muscle force decreased by 10–15% in the vastus and by 30–60% in the gluteus minimus (GMi) between 30% and 60% of the gait cycle (stance phase).

Effect of modification of the muscle excitation pattern (variant MII)

A modification of the muscle excitation pattern in accordance with Eq. (1) leads to the possibility of additional activity at the time when the muscle is elongated. This phenomenon can be seen using the example of rectus femoris muscle (Fig. 3). Between 3% and 54% of the gait cycle (marked with a frame),
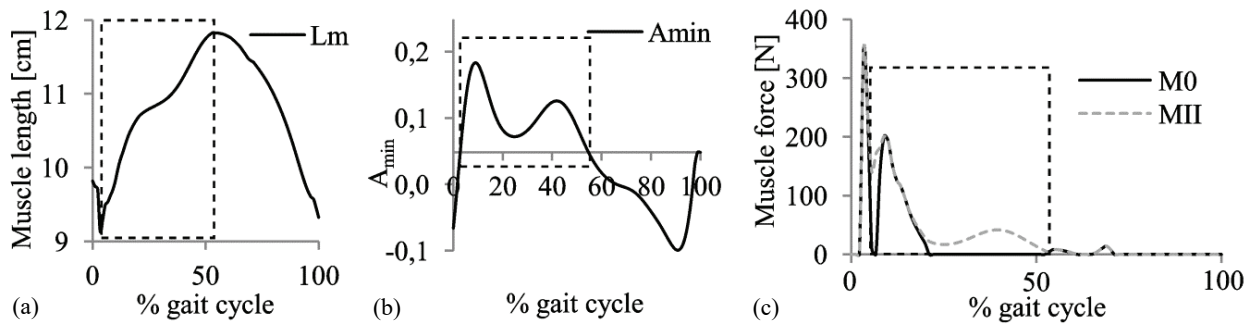
Fig. 3. The effect of the function used to change the pattern of muscle activity based on the example of rectus femoris depending on its elongation: (a) change in the muscle length (L_m); (b) the value of the function A_{min}; (c) active muscle force
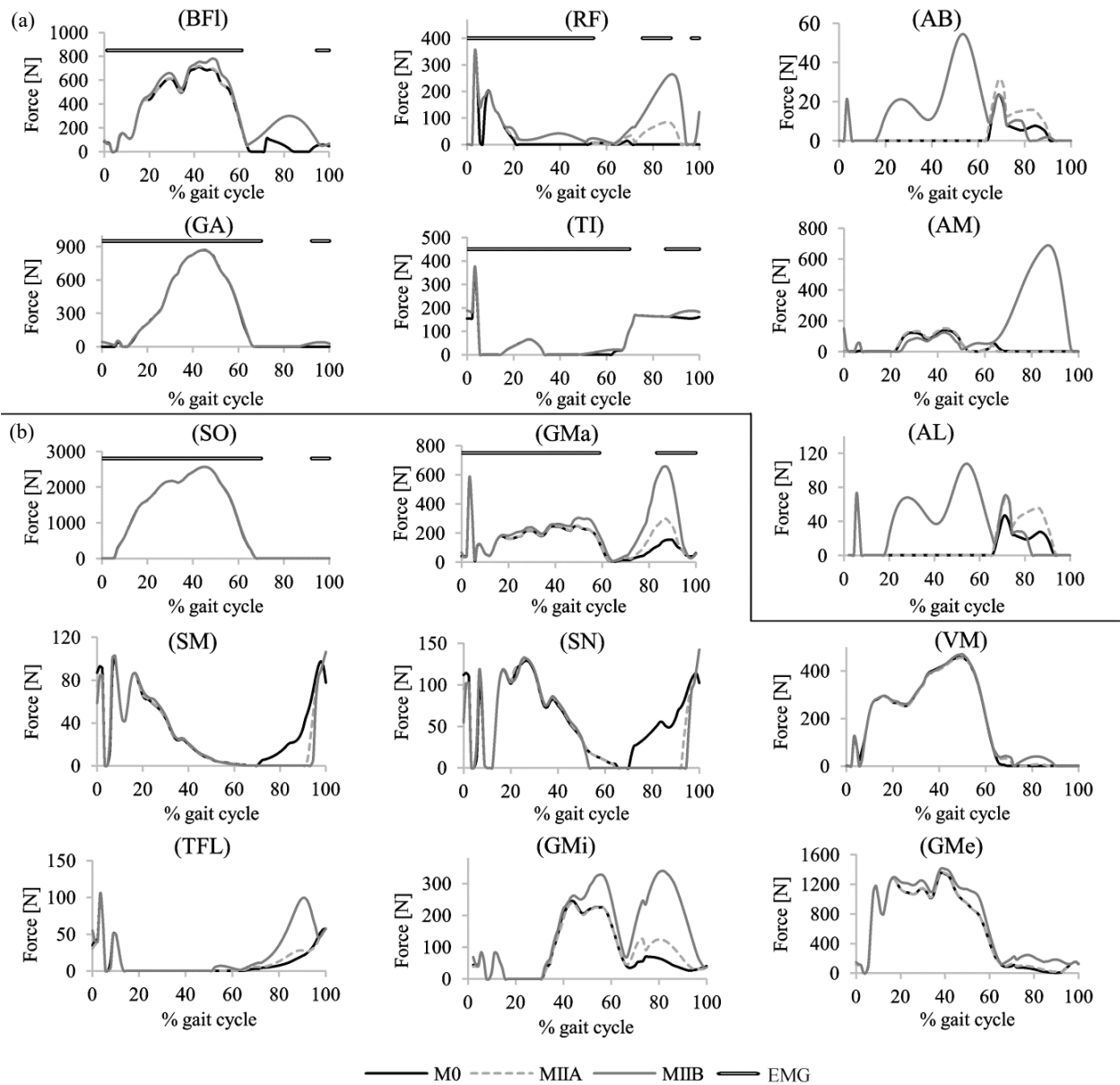

Fig. 4. Muscle active force waveforms for the muscles in which activity pattern of four muscles tested using the EMG (variant MIIA) and additional three hip adductors (variant MIIB) were modified (a) and for muscles without any changes in their models (b).
The results obtained for 2 variants of a model modification (MIIA, MIIB) are presented with reference to the results obtained for the standard model (M0). Abbreviations of the muscles as defined in the text
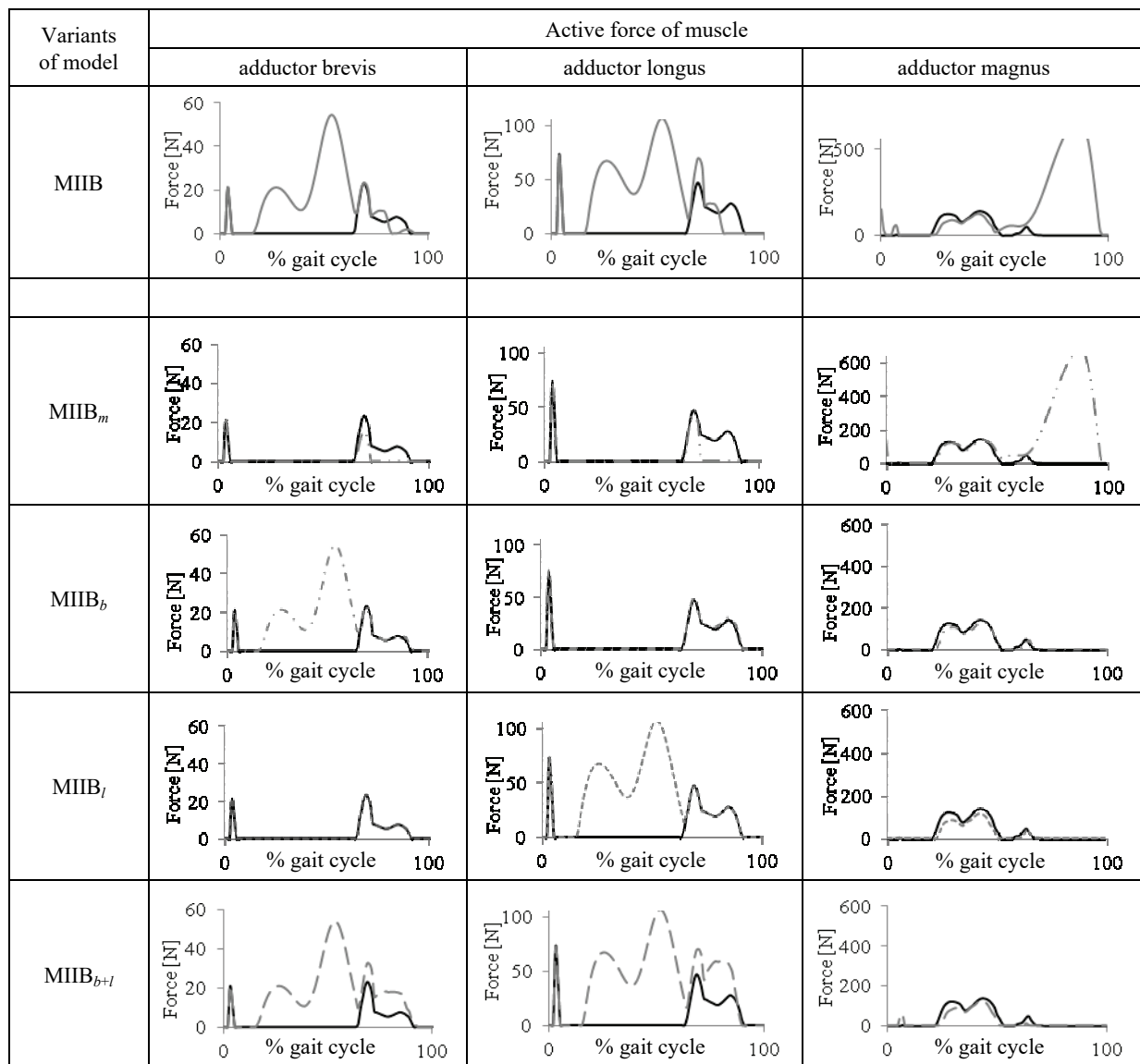
it may be observed that its length increases (Fig. 3a). In this situation, with a modified pattern of activity the lower muscle excitation boundary reaches a value above zero (parameter A_{min} – Fig. 3b), which results in the occurrence of a period of additional activity and of active muscle force generated as a consequence (Fig. 3c). This effect can be observed between 22 and 52% of the gait cycle. Another effect of modification of the excitation model may be an increase in active force, which takes place when the forced minimum excitation A_{min} is higher than the value obtained in the standard model (19–21% of gait cycle). When a muscle is shortened (0–3% and 54–100%), the velocity of change in the muscle length and consequently the lower boundary of the excitation A_{min} reach the values below zero, which does not cause any changes

in the muscle active force compared with the standard model (M0).

The results for the muscles in which the activity pattern of the muscle model was changed (change of four muscles tested using the EMG (Variant MIIA) and additionally three hip adductors (Variant MIIB) are presented in Fig. 4a. Activity period for particular muscles tested using the EMG was added for comparison. Results for the muscles whose model was not modified were presented in Fig. 4b.

The biggest influence of modification of the muscle excitation pattern (Variant MIIA) was observed in such actons which increase their length in the gait cycle and their activity generated in the case of the standard model was low or the muscle was inactive at all. Occurrence of a period of additional activity in all

Table 3. The effect of change in activity of adductor muscles on the muscle force patterns

Variants of model	Active force of muscle		
	adductor brevis	adductor longus	adductor magnus
MIIB			
$MIIB_m$			
$MIIB_b$			
$MIIB_l$			
$MIIB_{b+l}$			

——— M0 ——— MIIB —·—·— MIIBm —·—·— MIIBb —————— MIIBl — — MIIBb+l.

modified muscles (rectus femoris (RF), gastrocnemius (GA), tibialis anterior (TI) and biceps femoris caput longum (BFl)) was observed. Such results are closer to EMG recordings in the case of RF and GA. The modification of muscle activity patterns of tibialis anterior (TI) and biceps femoris caput longum (BFl) does not fully reflect the EMG recordings. It is associated with the fact that elongation patterns of this muscle do not fit to the taken formula of muscle excitation. Activity of tibialis anterior calculated with use of standard model (M0) did not reflect EMG recordings. However, this acton was not elongated during the major part of stance phase and its additional activity was obtained only for the final part of stance phase

(50–65% of gait cycle). Additional activity of biceps femoris caput longum calculated during swing phase does not reflect EMG recordings. It is possible that qualification of this muscle to the group of muscles with abnormal activity pattern was improper. Modification of the muscle excitation pattern for selected actons forced the changes of active force values generated in other muscles, including those whose model was not modified (Fig. 4b).

Adding adductors of the hip joint to the muscle group with a modified excitation pattern (Variant MIIB) results in the occurrence of periods of their additional activity. A detailed analysis of such phenomenon is presented in Table 3.
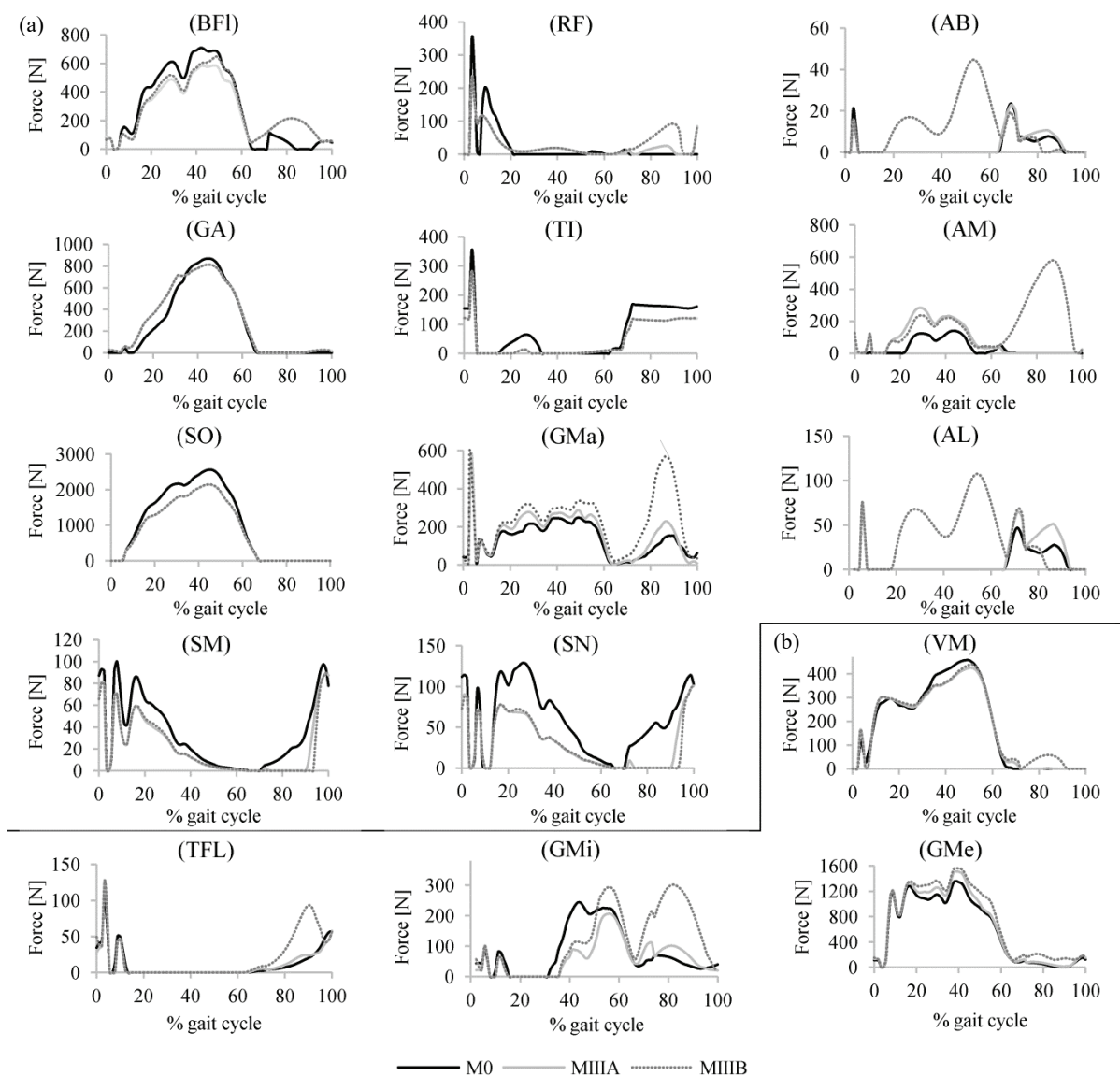

Fig. 5. Muscle active force waveforms for the muscles in which morphological parameters and/or activity pattern of four muscles tested using the EMG (variant MIIIA), and additional three hip adductors (variant MIIIB) were modified (A) and for muscles without any changes in their models (B).
The results obtained for 2 variants of a model modification (MIIIA, MIIIB) are presented with reference to the results obtained for the standard model (M0). Abbreviations of the muscles as defined in the text

When the way of activating only one of the adductors was respectively modified: adductor brevis (MIIB$_b$), adductor longus (MIIB$_l$), adductor magnus (MIIB$_m$), the occurrence of an additional period of active force generation in the muscle whose excitation pattern was modified with a decrease (or complete disappearance) of the muscle force in two remaining adductors was observed. The modification of the activity of two adductors together (MIIB$_{b+l}$) led to an increase in the AB and AL activity and a decrease in the AM activity in the stance phase. As a result of the simulation of pathological activity of all adductors (variant MIIB), their summary force increased, which had an influence on the behaviour of the whole group of muscles acting on the hip joint. The values of the forces generated by the biceps femoris and vastus muscle groups were higher in comparison with the results obtained in previously described variants of the model. However, the biggest effect was observed in the abductors of the hip joint (GMi, GMe, TFL). The value of the force in the gluteus minimus (GMi) increased from 40% of the gait cycle, and the active force in the swing phase was even 2–3 times higher compared to other modification variants. Moreover, the active force of gluteus medialis (GMe) in the stance phase (except the initial double stance – 0–19%) increased by 10–36% in comparison with the standard model and also increased in the swing phase (Fig. 4b).

Effect of modification of the muscle morphology and excitation pattern (variant MIII)

In the case of combined modification of the muscle morphology and excitation pattern (variant MIII), a significant influence of a change in morphological parameters on the active force value was observed.

The results for the muscles in which the model was changed in any way (change of morphological parameters and/or activity patterns) for two variants (MIIIA, MIIIB) are presented in Fig. 5a. Analogous results for the selected muscles whose model was not modified are presented in Fig. 5b.

The muscles in which the length of the contractile part and the volume of the belly were reduced and the minimum muscle excitation boundary was changed produced less force than in the case of modification where only the excitation pattern was changed (MII). This probably results from reduced morphological muscle parameters and, as a consequence, leads to lower active force. A comparison of the two variants of the combined modification of geometry and excitation pattern (variants MIIIA and MIIIB) revealed that in most cases (e.g. gluteus group, vastus group, adductor longus, rectus femoris) the muscle force was higher with a modification of the activity of hip adductors (variant MIIIB). This points to a huge impact of additional, abnormal activity of adductors of the hip joint on the whole group of lower limb muscles.

3.2. Analysis of the joint reaction forces in the hip

As a result of numerical studies, the joint reaction forces acting in the hip were determined. The value of the resultant force is presented in Fig. 6.

The reaction value obtained in all variants of the analysis is significantly higher than in normal locomotion (according to the research by Bergmann et al. 300–350% BW [4]) and can even reach 850% BW.
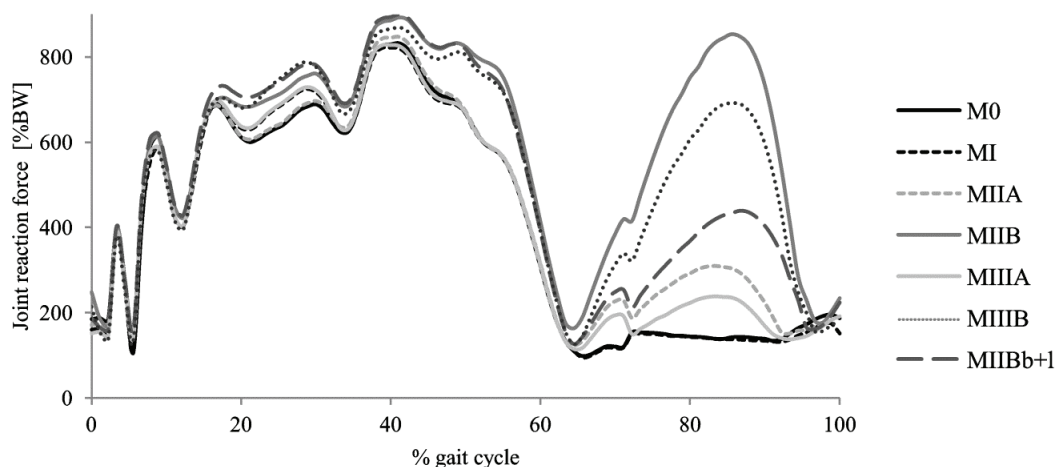

Fig. 6. The resultant hip-joint reaction force expressed as a percentage of the body weight.
The results for an analysis of various variants of the model modification (MI, MIIA, MIIB, MIIB$_{b+l}$, MIII, MIIIB)
are presented with reference to the results obtained for the standard model (M0)

In the case of modification of the model in the MI variant a reduction of morphological parameters had only a negligible influence on the value of resultant reaction forces acting on the hip joint. A reduction of the reaction force value of only several percent during the double stance phase and a slightly higher (7–10%) increase of the reaction force in the single stance phase (20–35% of the gait cycle) can be seen. The modification of the excitation pattern (variant MIIA) did not cause significant changes in the reaction value during the stance phase, however, the reaction values increased two times in the swing phase. The biggest impact on the value of the reaction force had the addition of adductors to a group of modified activity pattern (variants MIIB and MIIIB) where the resultant force increased by 3–6% in the double stance phase (17%–20% of the gait cycle) and by 5–24% in the limb single stance phase (28%–9% of gait cycle). From 63% of the gait cycle and during the swing phase, very high values of joint reaction forces whose maximum is lower only by 2–7% than the maximum of the reaction force values for the stance phase can be observed. It should be added that in the case of a change in excitation of the hip adductor muscles with the omission of AM (variant MIIB$_{b+l}$) the reaction during the single stance phase had similar values as with the modification of all adductors but was considerably lower in the swing phase. The results presented indicate a significant impact of abnormal functioning of the hip adductor on the reaction force acting on the hip joint.

4. Discussion

The subject of research described in this study was the use of a model of the musculoskeletal system for the analysis of the gait of a 14-year-old girl with cerebral palsy and with the symptoms of spastic diplegia. The methodological limitations due to the lack of complete data (e.g., MRI/ULTRASOUND) on the evaluated patient made it necessary to obtain the information about the geometry and the function of spastic muscles from the literature data. The information included in the clinical report suggested only which groups of muscles might (but do not have to) work incorrectly. The implementation of several variants of the standard muscle model modification (morphological structure and/or pattern of excitation) made it possible to assess the influence of potential pathological changes on the biomechanical parameters (muscle force, joint reaction force) during pathological gait of

the CP patient. The obtained results of the numerical analysis showed that the inclusion of incorrect excitation and different morphological structures of the selected muscles in the model had a significant influence on the results of the simulation of function of the entire musculoskeletal system as it comprised not only the muscles changed as a result of the disease.

First of all, the modification of morphological parameters caused a reduction of the value of active force generated by muscles with a changed geometry. A confirmation of the validity of this observation can be found in the studies by Lampe [15] and Barret [2], who indicated that the consequence of the structural deficits is primarily the reduced ability to generate the muscle force. It is worth noting that in the case of the gastrocnemius, gluteus maximus and adductor magnus muscles, despite a reduction of the morphological parameters, their active force increased as a result of large deficits of muscles performing the same function which may indicate that the compensation occurred within the respective muscle group (flexors or extensors). The reduction of parameters of the semitendinosus, semimembranosus, biceps femoris caput longum muscles was much higher than the reduction of parameters of the gluteus maximus and adductor magnus muscles, which could have been the major stimulator of the ensuing compensation within the muscles acting on the hip joint. However, when analysing the muscle function in the case of the crouch gait, it can be concluded that with the excessive flexion of the knee joints there are also changes in the moment arm of specific forces, which, in turn, makes the muscle group at the back of the thigh work mainly as knee flexors and not as hip extensors [9]. Considering the above and the large deficits of the muscles at the back of the thigh which contribute to an additional weakness of these muscles, it can be assumed that, as a consequence, the force generated by other extensors is increased in order to maintain the balance and the upright posture. It is obvious that the modification of morphological parameters of a certain muscle group (in this case, 10 actons) forced a change in the values of active forces generated in other muscles, not changed by the disease, and thus it affected the function of the entire musculoskeletal system.

In addition to the morphological parameters of muscles being changed, the literature indicates the abnormal muscle response to stretching [6], [14]. The modifications in which the incorrect muscle activity and excitation pattern were simulated (variant MII) resulted above all in the occurrence of an additional activity at the time when the muscle was elongated. In the case of modification of the muscle excitation pat-

tern measured using the EMG (TI, BF, RF, GA) the value of the force in the stance phase generally increased compared to the standard model. This effect is especially well visible in the swing phase when a two-fold increase in the muscle force was visible. A similar relationship in the studies on spastic lower leg muscle function was shown by Krogt where he even observed a threefold increase in the muscle activity in the swing phase [14]. However, a comparison of calculated activity of particular muscles with EMG recording showed some inaccuracy (e.g., improper activity of biceps femoris). It means that further research to elaborate criteria for selection of muscles with modified activity pattern must be conducted. Similarly, further analyses are necessary to find optimal value of the proportionality coefficient used in formula joining muscle excitation with its elongation velocity.

A very strong adduction of the hip combined with an excessive flexion of the knee and hip joints caused difficulties in carrying out the electromyography tests of the medial thigh muscles. However, the patient's posture described and the information about the crouch gait available in the literature [1], [19] suggest an incorrect activity of the hip adductors. For this reason, other variants of the muscle modification have been considered in order to additionally analyse the influence of incorrect stimulation of the hip adductors. The numerical simulations allowed us to consider several variants of any incorrect activity of the adductor muscles (only one, two or three adductors). The most interesting results were observed when all adductors showed an incorrect activity (variant MIIB) because an increase in the active force of each of them, but in different periods, was noticed (adductor brevis and adductor longus muscles in the stance phase and adductor magnus in the swing phase). Moreover, the influence of the MIIB modification was observed in the function of the entire musculoskeletal system and the greatest effect was noticed in the activity of the hip abductors. A simultaneous increase in the force of adductors and abductors may indicate the co-contraction of the analysed muscle groups being a characteristic phenomenon for spastic muscles [19], [22]. The results of an analysis including a combined modification of the muscle morphology and excitation pattern (variant MIII) clearly point to the necessity of making extensive changes to the model. It may be observed that the decrease in muscle force compared to the variant where only the muscle activity was changed (MII) was associated primarily with the reduction of morphological parameters. In addition, the results of analyses conducted showed very high values

of the reaction force acting in the hip compared to normal locomotion [4]. This effect can be partly associated with the well known fact that reaction forces calculated using musculoskeletal model tend to be overestimated with regard to the measured forces and they can even reach 350–400% BW [28]. Moreover, Miller [19] explained that the very high value of the reaction force in the spastic hip could be caused by the muscle co-contraction (hip flexors, hamstrings, adductors, and abductors) which might generate much higher forces than in the case of a healthy hip. For example, the total force of the hip joint adductors in the single stance phase calculated on the basis of our study has values 4 times the body weight of the patient evaluated, while in the correct gait this force equals approximately 2 times the body weight. Additionally, the emergence of very large muscle force values in the swing phase resulted in an excessive value of the reaction force acting on the hip joint. The inclusion of the adductors in the modified activity pattern group (variants MIIB and MIIIB) has a definitely strongest influence on the reaction value where an increase in the value of the resultant force in the stance phase and very high values of the reaction force in the swing phase are visible. It is noteworthy that in the case of only two adductors having an incorrect activity (variant MIIB$_{b+l}$) without the AM (which could be working mainly as an extensor) the values in the swing phase would have been much lower. The numerical simulation results presented indicate a significant influence of the modified morphological muscle parameters and excitation pattern on the function of the musculoskeletal system during the spastic gait. It seems advisable to conduct further research which would allow realistic modelling of the spastic muscles in a given patient (range of morphological changes, pathological activity). Additionally, an important aspect should also be a realistic reconstruction of existing bone deformities which might affect the conditions of equilibrium of the muscular forces within the analysed joint.

5. Conclusions

The results of the present study point to the advisability of realistic modelling of the pathologically changed muscles because it has a significant influence on the results of biomechanical analysis of the musculoskeletal system in cerebral palsy patients. It is also necessary to develop an optimal algorithm to determine the muscle model parameters depending on the

severity of the disease and the patient's individual characteristics.

Acknowledgement

References

[1] ARNOLD A.S., ASAKAWA D.J., DELP S.L., *Do the hamstrings and adductors contribute to excessive internal rotation of the hip in persons with cerebral palsy?*, Gait Posture, 2000, 11, 181–190.

[2] BARRETT R.S., LICHTWARK G.A., *Gross muscle morphology and structure in spastic cerebral palsy: a systematic review*, Dev. Med. Child. Neurol., 2010, 52(9), 794–804.

[3] BAX M., GOLDSTEIN M., *The Definition and Classification of Cerebral Palsy*, Dev. Med. Child. Neurol., 2007, 49, 144–151.

[4] BERGMANN G., DEURETZBACHER G., HELLER M., GRAICHEN F., ROHLMANN A., STRAUSS J., DUDA G.N., *Hip contact forces and gait patterns from routine activities*, J. Biomech., 2001, 34(7), 859–871.

[5] DESAILLY E., *Simulation of muscle retraction in Cerebral Palsy (SiMusCP). Validation of a decision support system for surgical lengthening of contractured muscles*, Comput. Method Biomech., 2012, 15, 263, 265.

[6] DE FREITAS T., *Muscle torque of healthy individuals and individuals with spastic hemiparesis after passive static streching*, Acta Bioeng. Biomech., 2016, 18(1), 35–39.

[7] DZIUBA A.K., *Index of mechanical work in gait of children with cerebral palsy*, Acta Bioeng. Biomech., 2014, 16(3), 77, 87.

[8] DZIUBA A., *Integral method (IM) as a quantitative and objective method to supplement the GMFCS classification of gait in children with cerebral palsy (CP)*, Acta Bioeng. Biomech., 2013, 15(2), 105, 111.

[9] GAGE J.R., *Gait Analysis: Principles and Applications*, J. Bone Joint Surg. Am., 1995, 77, 1607–1623.

[10] HAINISCH R., GFOEHLER M., ZUBAYER-UL-KARIM M., PANDY M.G., *Method for determining musculotendon parameters in subject-specific musculoskeletal models of children developed from MRI data*, Multibody Syst. Dyn., 2012, 28(1), 143–156.

[11] HANDSFIELD G.G., MEYER B.H., ABEL M.F., BLEMKER S.S., *Heterogeneity of muscle sizes in the lower limbs of children with cerebral palsy*, Muscle Nerve, 2016, 53(6), 933–945.

[12] HEINEN F., LUND M.E., RASMUSSEN J., DE ZEE M., *Muscle-tendon unit scaling methods of Hill-type musculoskeletal models: An overview*, P I MECH ENG H, 2016, 230 (10), 976–984.

[13] JONKERS I., *Musculo-tendon length and lengthening velocity of rectus femoris in stiff knee gait*, Gait Posture, 2006, 23, 222–229.

[14] KROGT M., *Dynamic spasticity of plantar flexor muscles in cerebral palsy gait*, J. Rehabil. Med., 2010, 42, 656–663.

[15] LAMPE R., GRASSL S., MITTERNACHT J., GERDESMEYER L., GRADINGER R., *MRT measurements of muscle volumes of the lower extremities of youths with spastic hemiplegia caused by cerebral palsy*, Brain Dev., 2006, 28, 500–506.

[16] LANCE J.W., *What is spasticity?*, Lancet, 1990, 335, 606.

[17] LEIGHTON R.D., *A functional model to describe the action of the adductor muscles at the hip in the transverse plane*, Physiotherapy Theory and Practice, 2006, 22(5), 251–262

[18] MALAIYA R., MCNEE A.E., FRY N.R., EVE L.C., GOUGH M., SHORTLAND A.P., *The morphology of the medial gastrocnemius in typically developing children and children with spastic hemiplegic cerebral palsy*, J. Electromyogr. Kines., 2007, 17, 657–663.

[19] MILLER F., *Cerebral palsy*, Springer, 2006, ISBN 0-387-20437-7.

[20] MOHAGHEGHI A.A., KHAN T., MEADOWS T.H., GIANNICAS K., BALTZOPOULOS V., MAGANARIS C.N., *Differences in gastrocnemius muscle architecture between the paretic and nonparetic legs in children with hemiplegic cerebral palsy*, Clin. Biomech., 2007, 22, 718–724.

[21] ODDING E., ROEBROECK M.E., STAM H.J., *The epidemiology of cerebral palsy: incidence, impairments and risk factors*, Disabil. Rehabil., 2006, 28(4), 183–191.

[22] OSKOUI M., COUTINHO F., DYKEMAN J., JETTE N., *An update on the prevalence of cerebral palsy: a systematic review and metaanalysis*, Dev. Med. Child. Neurol., 2013, 55(6), 509–519.

[23] SHEEHAN F.T., ZAJĄC F.E., DRACE J.E., *Using cine phase contrast magnetic resonance imaging to noninvasively study in vivo knee dynamics*, J. Biomech., 1998, 31, 21–26.

[24] SHORTLAND A.P., HARRIS C.A., GOUGH M., ROBINSON R., *Architecture of the medial gastrocnemius in children with spastic diplegia*, Dev. Med. Child. Neurol., 2002, 44, 158–163.

[25] SYCZEWSKA M., ŚWIĘCICKA A., *Are electromyographic patterns during gait related to abnormality level of the gait in patients with spastic cerebral palsy?*, Acta Bioeng. Biomech., 2016, 18(3), 91, 96.

[26] VAN DER KROGT M.M., BAR-ON L., KINDT T., DESLOOVERE K., HARLAAR J., *Neuro-musculoskeletal simulation of instrumented contracture and spasticity assessment in children with cerebral palsy*, J. Neuroeng. Rehabil., 2016, 13, 64.

[27] DAMSGAARD M., RASMUSSEN J., SOREN T.C., SURMA E., DE ZEE M., *Analysis of musculoskeletal systems in the AnyBody modeling system*, Simul. Model Pract. Th., 2006, 14, 1100–1111.

[28] THIELEN T., MAAS S., ZÜRBES A., WALDMANN D., KELM J., *Development of a hip interim prosthesis (spacer) using finite element analysis taking into account the muscle and joint forces from AnyBody*, 2009, ANSYS Conference and 27th CADFEM Users' Meeting, http://prod.anybodytech.combineservices.dk/downloads/publications., Accessed 20 January 2017.