Approach for Determination of Functioning of Lower Limb Muscles

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Abstract The purpose of the study is elaboration of approach for determination of functioning of chosen muscles that are essential for gait performance (Tibialis Anterior, Rectus Femoris, Gastrocnemius Medialis, Biceps Femoris). The scope of the study involves the analysis of the symmetric planar motion performing in the sagittal plane of the body by applying planar multibody model and electromyography signals (EMG) registered over normal gait performance. The analysis is performed by applying two types of multibody model: six degree of freedom system and seven degree of freedom system. Inverse dynamics task was used to calculated joint moments influenced ankle joints, knee joints and hip joints. Applied model also described single support phase and double support phase by taking into consideration the model of interaction between the ground and the contact foot. The activity states of considered muscles are determined on the base of their average activations and sequences in time.

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1 Introduction

The gait is a complex performance, which is induced by nervous system function (due to reflexes), muscle system action, skeletal system and external load configuration including gravitational field influence. Brain induces muscle activation that provide to gait performance (physical activity of a person) and muscle activation depends on descending and reflex inputs [7].

To estimate activity of muscles inducing a motion the electromyography (EMG) system is used [14] and muscle excitation timing can be estimated [8, 9]. To analyze EMG data there are applied processing algorithms, e.g. rectifying, smoothing, filtering, normalization [4]. Furthermore, one should consider the action of muscle by taking into account the path of this muscle and skeletal system configuration. It is worth noticing that the function of one-joint muscle is different from the function of two-joint muscle (or multi-joint muscle) [19].

To improve and stabilize movement activity a muscle co-contraction mechanism can occur [11]. During the prolonged action of this mechanism the metabolic transformation is growing up and this may increase the risk of damage of muscles causing this co-contraction action.

The aim of this study was to elaborate an approach for determination of functioning (activity) of the chosen lower limb muscles, which are essential for human gait performance. The scope of the study involves the analysis of the symmetric planar motion performing in the sagittal plane of the body by applying planar multibody model and EMG signals registered over normal gait performance.

2 Method

In the scope of this study it was investigated an influence of four muscles of lower limb: Tibialis Anterior (1), Rectus Femoris (2), Gastrocnemius Medialis (3), Biceps Femoris (4) (Fig. 1). These muscles were chosen due to the fact that they are superficial muscles that have a high priority in gait performance (the number of chosen muscles was caused by the limited possibility of EMG system used in experiments). The Tibialis Anterior is a one-joint muscle that performs dorsiflexion of the foot and slight inversion at the ankle joint [10]. The Rectus Femoris is a two-joint muscle that conducts flexion of the hip joint and extension of the knee joint. The Gastrocnemius Medialis is a two-joint muscle that performs plantar flexion of the foot at the ankle and flexion of the knee joint. The Biceps Femoris is a two-joint muscle that conducts: flexion and lateral rotation of the knee joint; extension and lateral rotation of the hip joint.

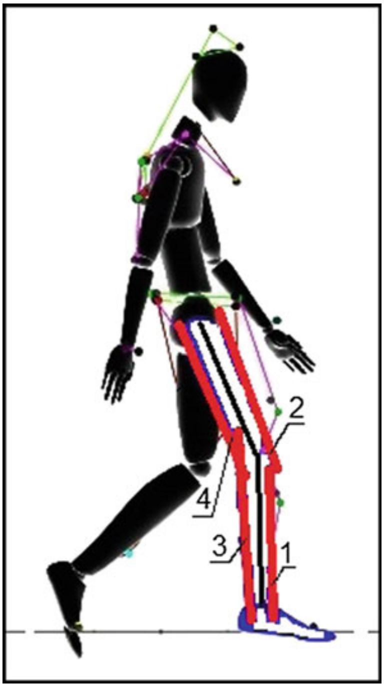

Fig. 1 Muscle system examined: Tibialis Anterior (1), Rectus Femoris (2), Gastrocnemius Medialis (3), Biceps Femoris (4)

To determine functioning of chosen muscles of lower limb over given gait phase the following inference method was developed. According to this method, one should first determine the condition of the joint examined [i.e. it is stiffened (blocked) or released (unblocked)]. After that one should consider whether a co-contraction of muscles acting on this joint is observed. To apply proposed inference method, one should obtain the following data:

(1) kinematic data (relative angular displacements, relative angular velocities and relative angular accelerations of lower limb segments);
(2) kinetic data (net joint moments (joint moments));
(3) EMG data (measured and processed EMG).

In order to determine whether considered joint is stabilized or is released, one should analyze kinematic data by considering the influence of gravitational force and skeletal system configuration (in what way skeletal components interact with each other). To consider a muscle co-contraction one should analyze muscle influence by taking into account the motor function of muscles. It was assumed that functioning (activity) of each considered muscle is determined by average activation and its sequence in time.

To perform a kinetic (dynamic) analysis of normal human gait in the single support and the double support phase, two biomechanical models were used [17]. There were developed by applying the Newton-Euler approach [1] and the method of seg-

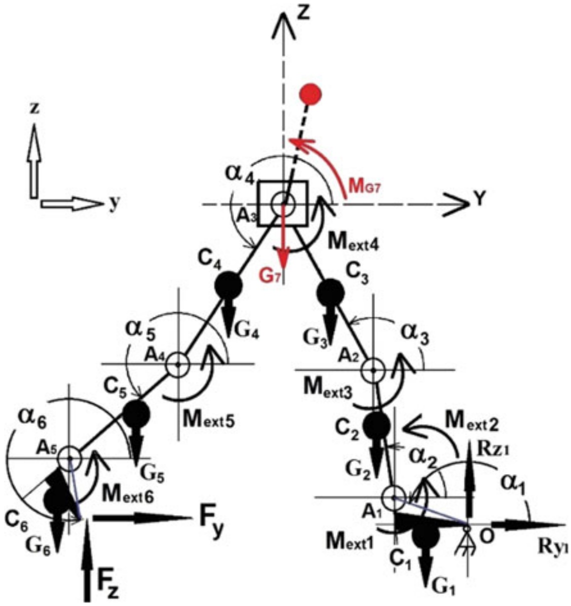

Fig. 2 The 6DOF model (O—the point between the support foot and the ground; A_1—the ankle joint of stance leg; A_2—the knee joint of stance leg; A_3—the hip joint; A_4—the knee joint of swing leg; A_5—the hip joint of swing leg; α_i—the angle of the i-th segment (each angle is measured as an absolute coordinate); G_i—the gravity force of the i-th segment that acts at its centre of gravity C_i; M_{exti}—the external moment acting on the i-th segment; F_y and F_z—the y-th component and z-th component of the reaction force influenced by the ground; R_{y1}—the y-th component of stance leg reaction force; R_{z1}—the z-th component of the stance leg reaction force; y—the sagittal axis; z—the vertical axis) [17]

mentation described in [2]. The first biomechanical model is the 6DOF model that treats a human body as a structure composed of six segments serially linked by hinge joints (Fig. 2). The second biomechanical model is the 7DOF model that treats a body as a dendritical structure composed of seven segments linked by hinge joints (Fig. 3). Over each phase of the gait the 6DOF or 7DOF model is connected to the ground at the joint O. During the single support phase two interaction forces acting at the sixth segment are equal to zero ($\underline{F_y} = 0$ and $F_z = 0$), whereas over the double support phase these two forces have some defined values ($\underline{F_y} \neq 0$ and $F_z \neq 0$). These values can be measured by applying a force plate (an external device used to measure interaction forces between the foot and the ground).

The 6DOF model assumes that a load of the upper part of the body (a force G_7 and the moment of this force M_{G7}) influences the stance leg. The mathematical form of the 6DOF model is a non-linear system of six equations:

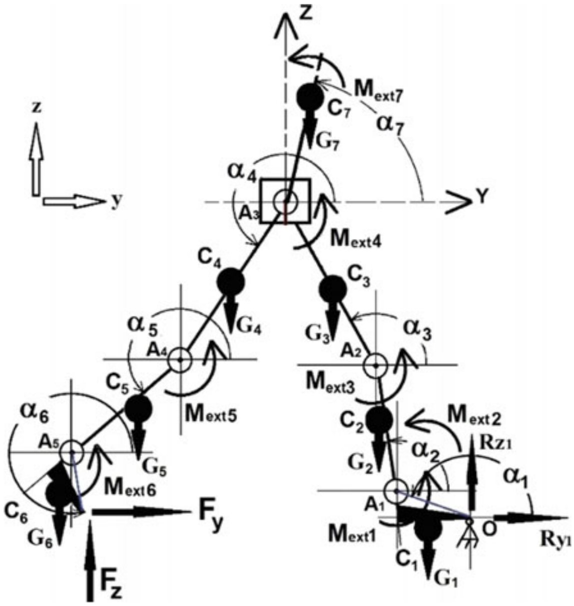

Fig. 3 The 7DOF model (symbols are described in the Fig. 2) [17]

$$
\begin{bmatrix}
A'_{11} & A'_{12} & A'_{13} & A'_{14} & A'_{15} & A'_{16} \\
A'_{21} & A'_{22} & A'_{23} & A'_{24} & A'_{25} & A'_{26} \\
A'_{31} & A'_{32} & A'_{33} & A'_{34} & A'_{35} & A'_{36} \\
A'_{41} & A'_{42} & A'_{43} & A'_{44} & A'_{45} & A'_{46} \\
A'_{51} & A'_{52} & A'_{53} & A'_{54} & A'_{55} & A'_{57} \\
A'_{61} & A'_{62} & A'_{63} & A'_{64} & A'_{65} & A'_{67}
\end{bmatrix}
\cdot
\begin{bmatrix}
\ddot{\alpha}_1 \\ \ddot{\alpha}_2 \\ \ddot{\alpha}_3 \\ \ddot{\alpha}_4 \\ \ddot{\alpha}_5 \\ \ddot{\alpha}_6
\end{bmatrix}
=
\begin{bmatrix}
M'_1(\alpha_1, F_y, F_z) \\
M'_2(\alpha_2, F_y, F_z) \\
M'_3(\alpha_3, F_y, F_z) \\
M'_4(\alpha_4, F_y, F_z) \\
M'_5(\alpha_5, F_y, F_z) \\
M'_6(\alpha_6, F_y, F_z)
\end{bmatrix}, \qquad (1)
$$

where: α_i—the i-th angular displacement of the i-th segment (the i-th joint angle); $\ddot{\alpha}_i$—the i-th angular acceleration of the i-th segment; $M'_i(\alpha_i, F_y, F_z)$—the i-th segment moment of the 6DOF model; A'_{ij}—non-linear functional coefficient of the 6DOF model that depends on the segment mass, segment length, segment radius of gyration, segment moment of inertia and segment angular displacements.

The 7DOF model assumes that the upper part of the body is modelled as one segment (the seventh segment). The mathematical form of the 7DOF model is a non-linear system of seven equations:

$$\begin{bmatrix} A''_{11} & A''_{12} & A''_{13} & A''_{14} & A''_{15} & A''_{16} & A''_{17} \\ A''_{21} & A''_{22} & A''_{23} & A''_{24} & A''_{25} & A''_{26} & A''_{27} \\ A''_{31} & A''_{32} & A''_{33} & A''_{34} & A''_{35} & A''_{36} & A''_{37} \\ A''_{41} & A''_{42} & A''_{43} & A''_{44} & A''_{45} & A''_{46} & A''_{47} \\ A''_{51} & A''_{52} & A''_{53} & A''_{54} & A''_{55} & A''_{56} & A''_{57} \\ A''_{61} & A''_{62} & A''_{63} & A''_{64} & A''_{65} & A''_{66} & A''_{67} \\ A''_{71} & A''_{72} & A''_{73} & A''_{74} & A''_{75} & A''_{76} & A''_{77} \end{bmatrix} \cdot \begin{bmatrix} \ddot{\alpha}_1 \\ \ddot{\alpha}_2 \\ \ddot{\alpha}_3 \\ \ddot{\alpha}_4 \\ \ddot{\alpha}_5 \\ \ddot{\alpha}_6 \\ \ddot{\alpha}_7 \end{bmatrix} = \begin{bmatrix} M''_1(\alpha_1, F_y, F_z) \\ M''_2(\alpha_2, F_y, F_z) \\ M''_3(\alpha_3, F_y, F_z) \\ M''_4(\alpha_4, F_y, F_z) \\ M''_5(\alpha_5, F_y, F_z) \\ M''_6(\alpha_6, F_y, F_z) \\ M''_7(\alpha_7) \end{bmatrix}, \qquad (2)$$

where: $M''_i(\alpha_i, F_y, F_z)$—the i-th segment moment of the 7DOF model; A''_{ij}—nonlinear functional coefficient of the 7DOF model that depends on the segment mass, segment length, segment radius of gyration, segment moment of inertia and segment angular displacements.

It is worth emphasizing that the i-th segment moment ($M'_i(\alpha_i, F_y, F_z)$ of the 6DOF model and $M''_i(\alpha_i, F_y, F_z)$ of the 7DOF model) depends on the joint moment M_{ij} generated between the i-th segment and j-th segment ($M_{ij} = M_{ji}$).

3 Application

Proposed inference method was applied to study the right lower limb having four EMG electrodes fixed according to the SENIAM requirement. Investigation was performed over one single support phase (that occurs after the toe-off of the left limb, i.e. from the 10% of stride) and following one double support phase to 58% stride period. To perform a normal gait analysis an experimental testing was conducted on one male health person [70.2 kg body mass, 183 cm body height, moments of inertia (Table 1)] that did five full steps (a middle step was taken into consideration). To obtain kinematic data a motion capture system OptiTrack Flex 13 with dedicated software was used. To measure interaction forces the Steinbichler force plate was applied. To measure surface EMG signals the Noraxon Myotrace 400 with MyoResearch XP Clinical Edition software was used (four channels system). The barefoot person walked with preferred speed in vision-on mode. It was assumed that all tests were done in a homogenic gravity field (with constant gravity acceleration).

Kinematic data were calculated by applying method presented in [15] and methods of signal processing (filtering, interpolating and differentiating): relative angular displacements are presented in Fig. 4; relative angular velocities are given in Fig. 5; relative angular accelerations are shown in Fig. 6.

Kinetic data (joint moments) were estimated by solving an inverse dynamics task and applying two biomechanical models (the 6DOF model and the 7DOF model). For this, kinematic data and measured interaction forces were used as input data. Kinetic

Table 1 Moments of inertia

Moment of inertia	6DOF model (kg m^2)	7DOF model (kg m^2)	Comment
J_1	0.0060	0.0060	The moment is calculated with respect to the p. O
J_2	0.2223	0.2223	The moment is calculated with respect to the p. A_1
J_3	1.1859	1.1859	The moment is calculated with respect to the p. A_2
J_4	0.6877	0.6877	The moment is calculated with respect to the p. A_3
J_5	0.1538	0.1538	The moment is calculated with respect to the p. A_4
J_6	0.0054	0.0054	The moment is calculated with respect to the p. A_5
J_7	–	3.5719	The moment is calculated with respect to the p. A_3

data calculated for 6DOF are presented in Fig. 7, whereas kinetic data calculated for 7DOF are shown in Fig. 8a, b.

In order to analyze EMG data measured there were rectified and normalized (to the maximum values) and processed by applying Root Mean Square algorithm (RMS) with the 10 ms frame (Fig. 9) and 50 ms frame (Fig. 10). The threshold of EMG was assumed to be equal 0.2 of the normalized RMS value.

To estimate the phases of joint stiffness it was taken into consideration that the threshold range of relative angular acceleration should be equal ± 1 rad/s^2 (Fig. 11).

4 Discussion

Considering kinematic data (Figs. 4, 5 and 6) and joint stiffness phases, (Fig. 11) it was defined that over the [10; 58]% of stride the right leg, which is the stance leg during the single support phase, was stabilized: (1) at the ankle joint over [22.5; 25.8]% of stride (stage IA) and [27.5; 38]% of stride (stage IB); (2) at the knee joint over [23; 33]% of stride (stage II); (3) at the hip joint over [24; 29] % of stride (stage III). It is worth noticing that given results are consistent with data described in [14]. Based on obtained results it was concluded that over the single support phase the

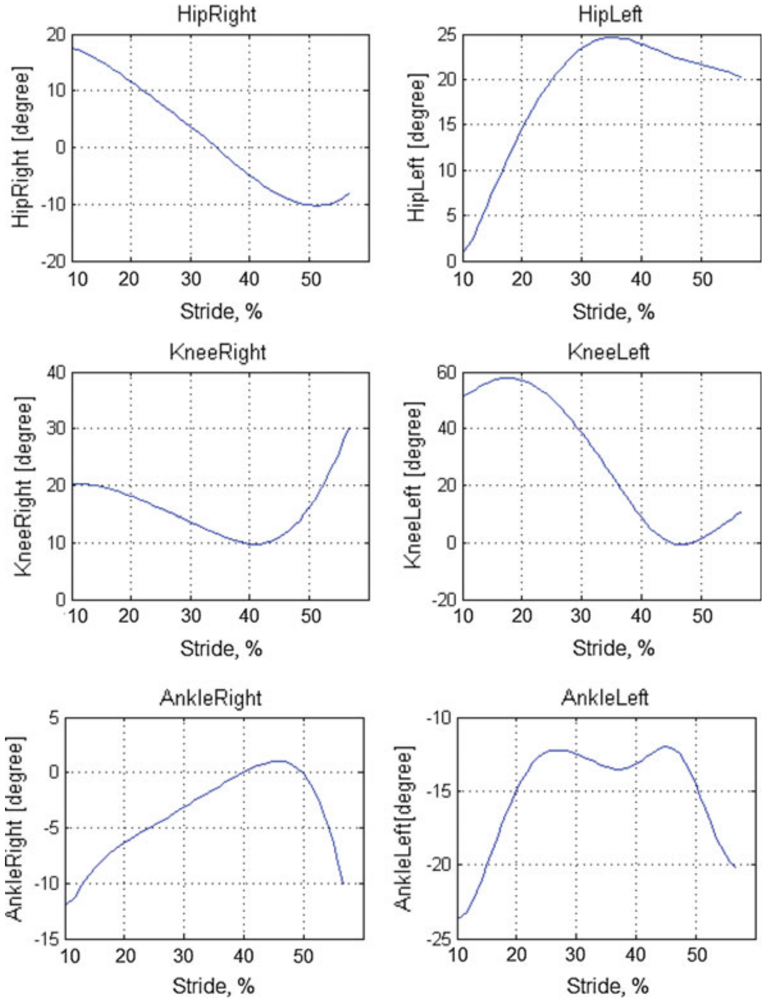

Fig. 4 Kinematic data: relative angular displacements

ankle joint of the stance leg should be stabilized as the first joint, the knee joint of this leg should be stabilized as the second joint and the hip joint of this leg should be stabilized as the third joint. After this the releasing of these joints should be performed in the reverse sequence: the hip joint should be released as the first joint, the knee joint should be released as the second joint and the ankle joint should be released as the third joint.

Analyzing kinetic data (joint moments) and joint stiffness phases (Fig. 11), it was noticed that: (1) over the stage IA and stage IB a joint moment calculated at the ankle joint ($M_{jointA1}$) is an increasing function (for the 6DOF model) or a wavy increasing function (for the 7DOF model); (2) over the stage II a joint moment calculated at

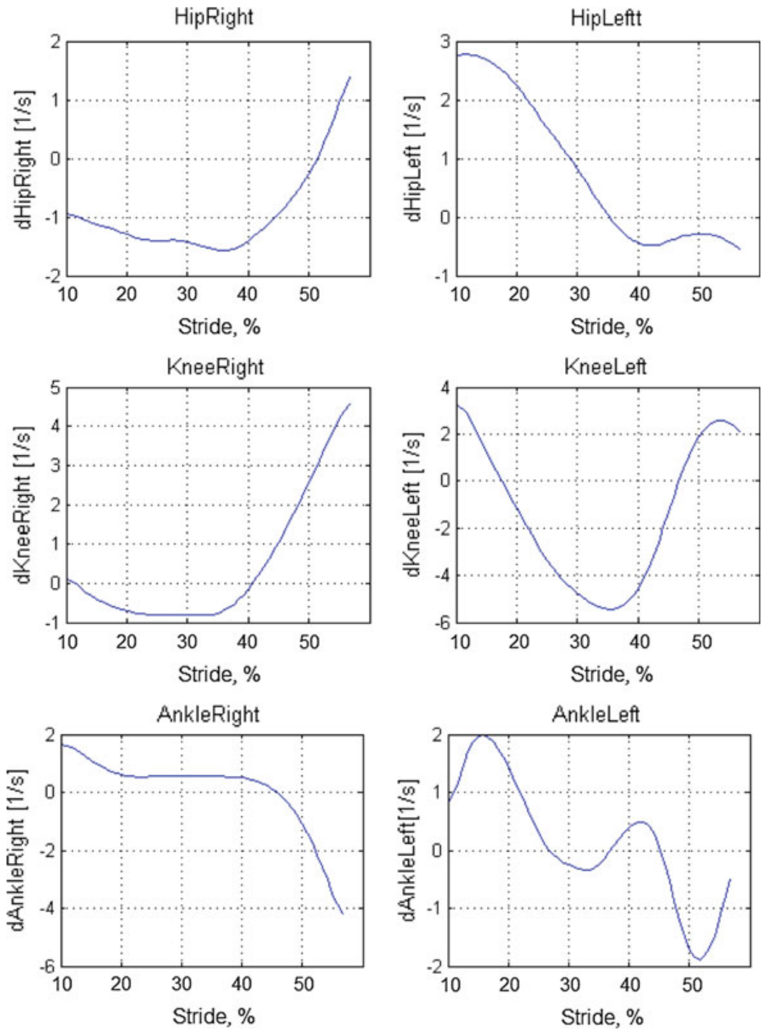

Fig. 5 Kinematic data: relative angular velocities

the knee joint ($M_{jointA2}$) is an increasing function (for the 6DOF model) or wavy function that changes its sign (for the 7DOF model); (3) over the stage III a joint moment calculated at the hip joint ($M_{jointA3}$) is equal to zero (for the 6DOF model and 7DOF model). It is worth emphasizing that calculated kinetic data only give us information whether the considered joint is loaded or unloaded. These data do not allow us to conclude whether the load at the joint is transmitted through antagonistic muscle pairs (muscle co-contraction), passive tissues or joint interaction (contact phenomena).

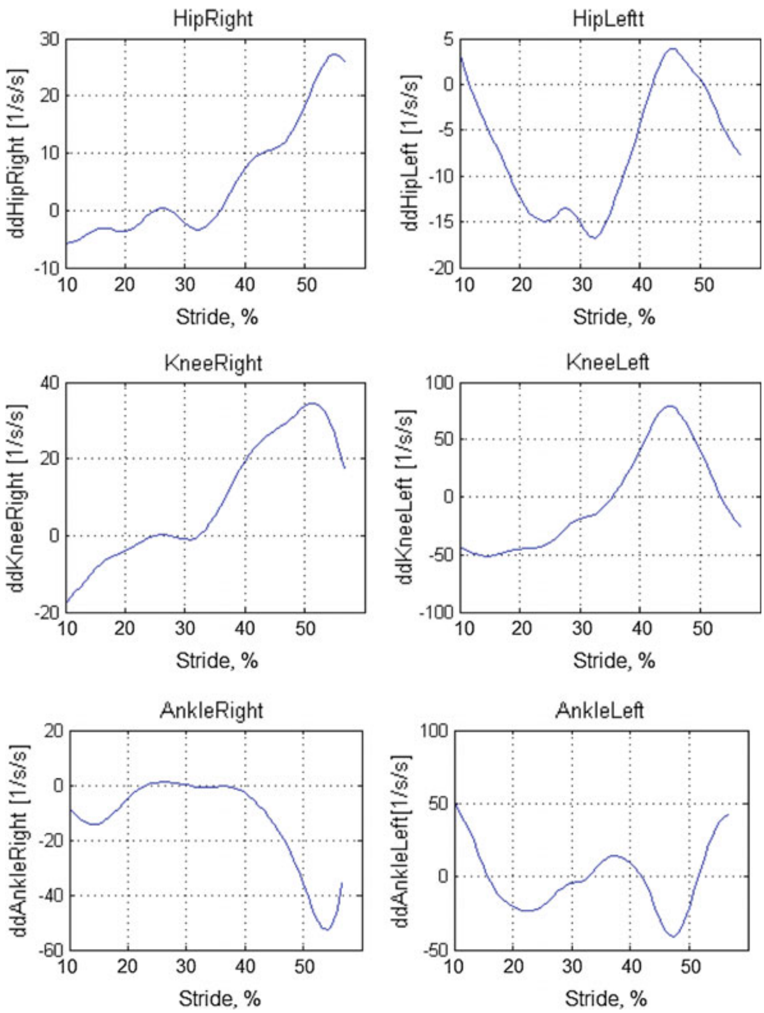

Fig. 6 Kinematic data: relative angular accelerations

Considering EMG data, muscle activity state of each examined muscle was determined. This state is described by the average activation time and its sequence in time. Analyzing normalized RMS EMG data (Figs. 9 and 10), average activation times were calculated:

(1) for the 10 ms frame RMS the activation times are equal 0.648 ms for the first (1), 0.293 ms for the second (2), 0.511 ms for the third (3) and 0.498 ms for the four muscles (4), respectively;

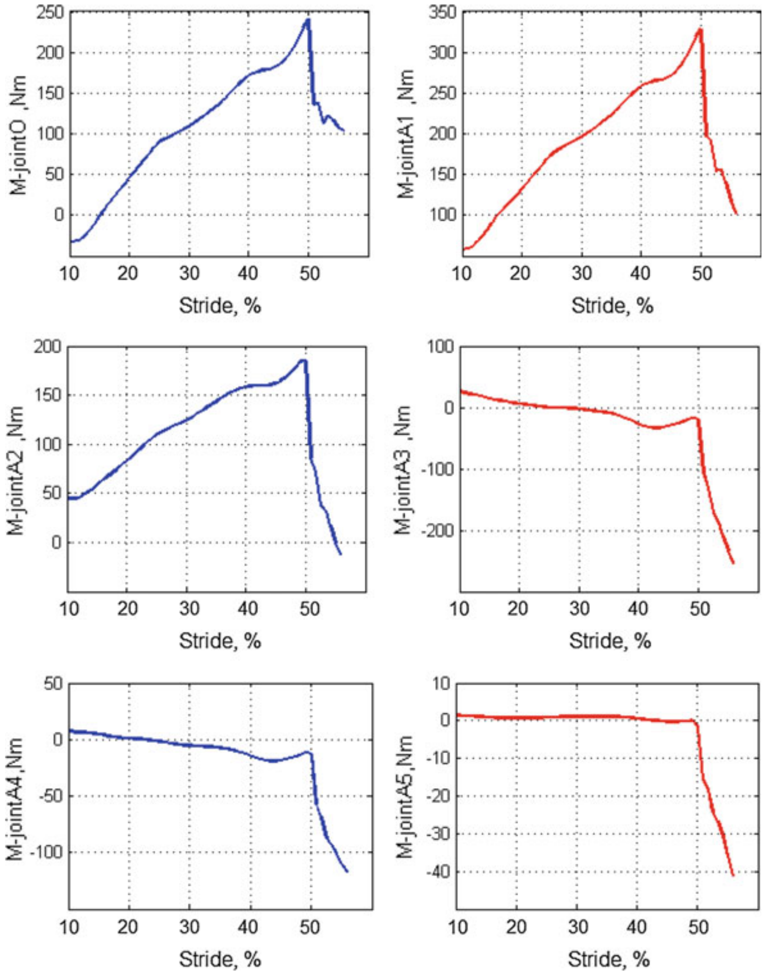

Fig. 7 Numerical results of 6DOF model: joint moments

(2) for the 50 ms frame RMS the activation times are equal 0.924 ms for the first (1), 0.526 ms for the second (2), 0.891 ms for the third (3) and 0.765 ms for the four muscles (4), respectively.

It is worth noticing that due to the fact that the processing algorithm has a huge impact on the data obtained (the 10 ms RMS method gives smaller values that the 50 ms RMS method) one should select a time frame on the base of the type of physical activity. In this study, we used the 50 ms RMS method because an examined walking performance can be treated as a moderate physical activity. Analyzing EMG processed data (Fig. 10) and joint stiffness phases (Fig. 11) it was observed that:

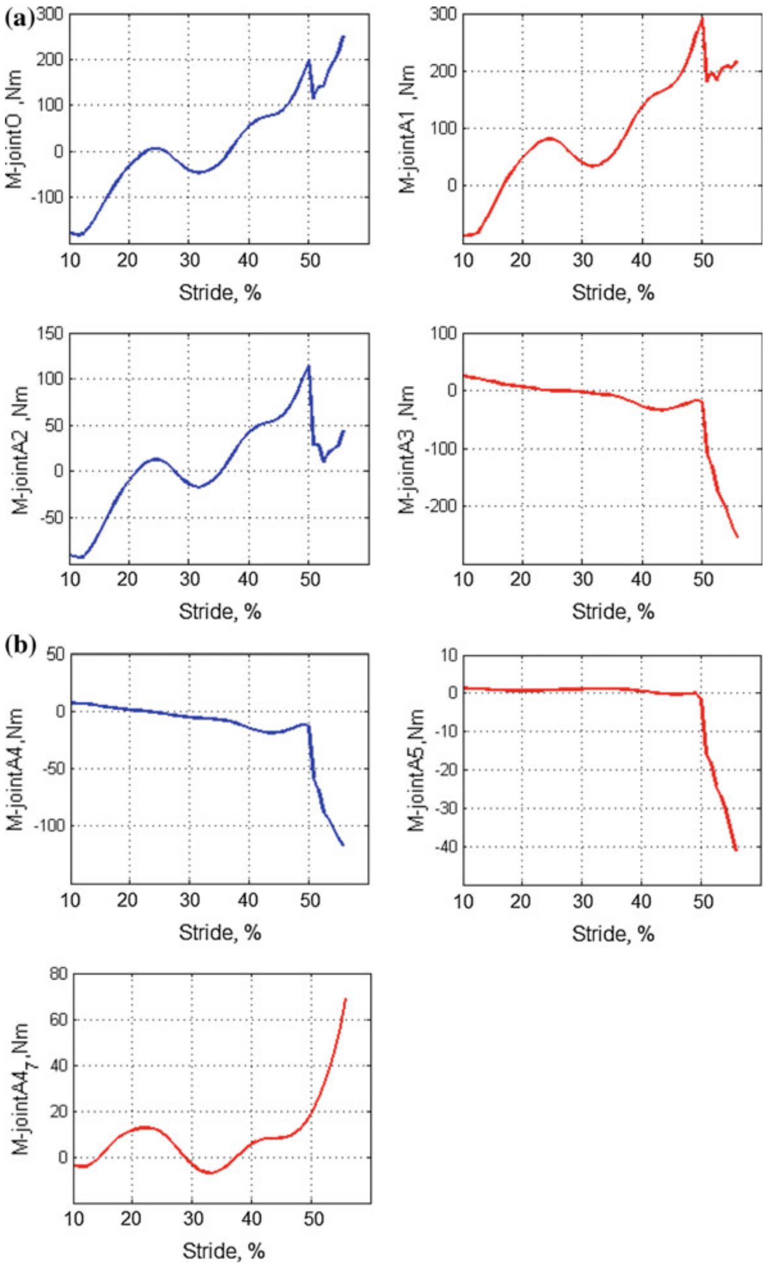

Fig. 8 A. Numerical results of 7DOF model: joint moments, B. Numerical results of 7DOF model: joint moments

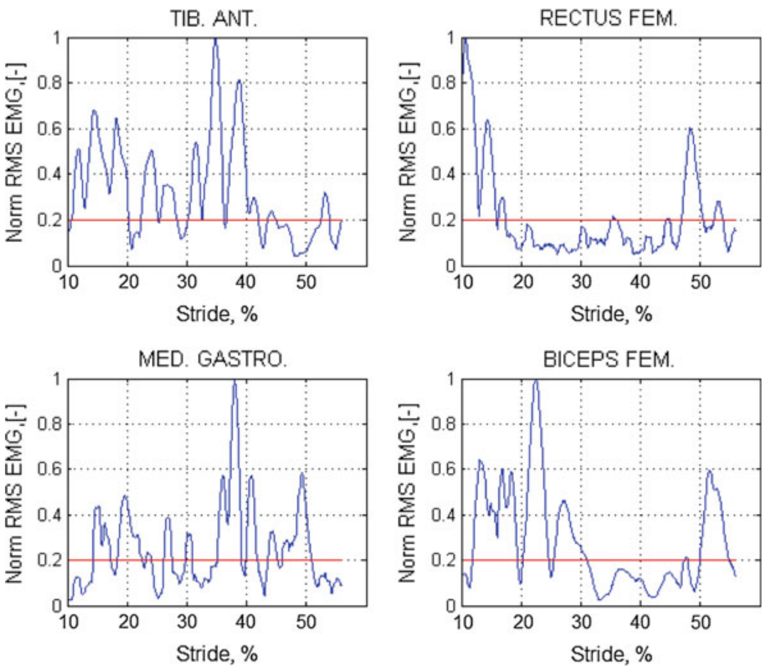

Fig. 9 Normalized RMS EMG signal (10 ms RMS)

(1) the antagonistic muscles acting on the ankle joint (the dorsiflexor muscle (1) and plantarflexor muscle (3)) showed similar activities over the stage IA and stage IB; moreover, over the second part of the stage IB it was noticed an occurrence of muscle co-contraction;
(2) the three muscles acting at the knee joint (the extensor muscle (2), the flexor muscle (3) and the flexor muscle (4)) presented different activities without any muscle co-contraction;
(3) the antagonistic muscles acting on the hip joint (the flexor muscle (2) and extensor muscle (3)) showed different activities over the stage III without any muscle co-contraction.

Considering the sequence of each muscle activity, it was noticed that obtained results are consistent with results presented in [14].

5 Conclusions

The aim of this study was elaborating an approach (inference method) that can be applied to determine functioning (activity) of lower limb muscles that are essential for gait performance over single support phase and double support phase by assuming

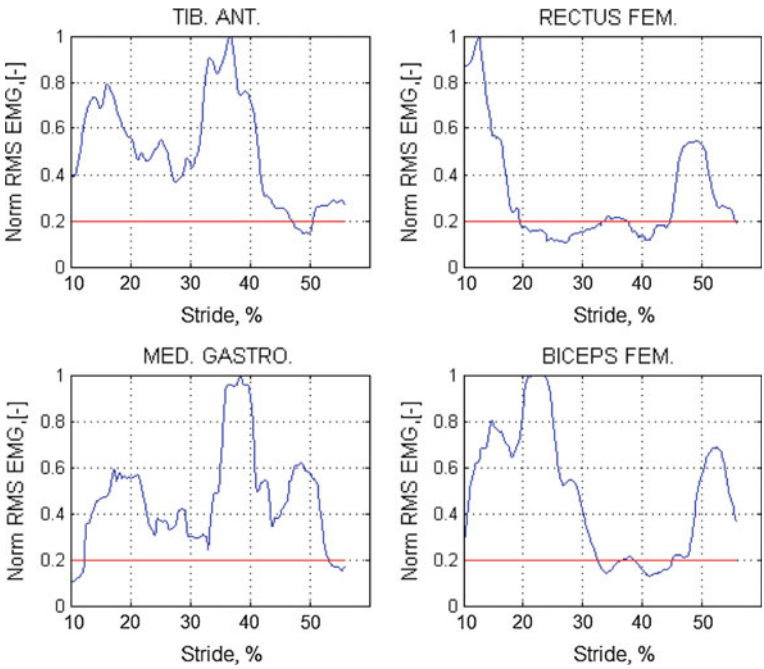

Fig. 10 Normalized RMS EMG signal (50 ms RMS)

the model of interaction between the ground and the contact foot. The scope of the study involved the analysis of the symmetric planar motion performing in the sagittal plane of the body by measuring kinematic, kinetic (force plate) and EMG data over normal gait performance. The kinetic (dynamic) analysis was performed by applying two multibody models (the 6DOF model and the 7DOF model) and solving an inverse dynamics task. It was assumed that activity state of each considered muscle is determined by average activation time and its sequence in time.

According to the presented approach, the conditions of the joint examined (stiffened or released) should be determined on the base of kinematic data by taking into consideration the influence of gravitational field and skeletal system configuration. After this one could consider processed EMG data and functions of examined muscles to conclude whether a muscle co-contraction was occurred. It is worth noticing that the method of EMG data processing has a huge impact on the result of muscle co-contraction investigation.

One should keep in mind that an inverse dynamics approach does not allow predicting an occurrence of muscle co-contraction [3]. The reason of this is the fact that during a co-contraction phase the agonist group and antagonist group produce moments that stabilize a joint at the same time. That is why calculated joint moment (net joint moment) does not give us any information what is the share of agonist group, what is the share of antagonist group, what is the share of passive tissues

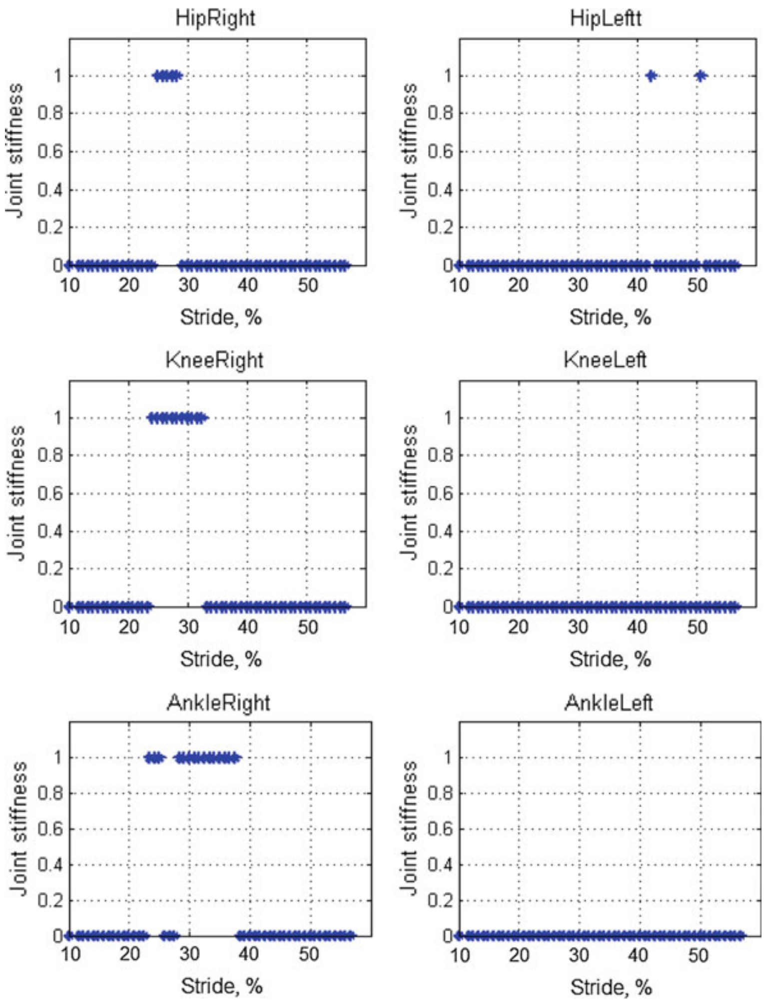

Fig. 11 Joint stiffness phases

influencing this joint and what is the share of contact forces acting at the joint. To detect a muscle co-contraction, one should measure and analyze EMG data of the muscles acting on the examined joint and consider kinematic data. It is worth emphasizing that information about an occurrence of muscle co-contraction is very important for clinical gait analysis and development of external device that helps to produce gait (e.g. exoskeleton for lower limb rehabilitation).

The future development of approach presented in this study involves: (1) analyzing influence of all superficial muscles that are essential for gait performance and can be measured by using surface EMG system (due to the fact that only non-invasive method can be used in the scope of presented study); (2) analyzing clinical gait performance on the base of an index quantifying deviation from normal gait [12, 13]; (3) considering muscle biomechanics [16, 18] and estimation of muscle synergy indices [5] to elaborate a method for solving the problem of redundancy [6].

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