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Compensatory strategy for ankle dorsiflexion muscle weakness during gait in patients with drop-foot

Michalina Błażkiewicz*, Andrzej Wit

The Józef Piłsudski University of Physical Education, Faculty of Rehabilitation, Marymoncka 34, Warsaw, 00-968, Poland

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ABSTRACT

Background: Pathological movement patterns are characterized by abnormal kinematics, kinetics and muscle activations that alter the distribution of muscle forces during walking.
Aim: The objective of this study was to identify what compensatory strategy is evident in muscle force distribution in patients with drop-foot, in response to weakness in the dorsiflexor muscles.
Methods: A sample of 10 patients with drop-foot were evaluated by a computerized gait analysis system and compared to a group of 10 healthy subjects. Muscle-actuated simulations of normal and drop-foot walking were performed using OpenSim software. A musculoskeletal model with 43 muscles acting on one lower extremity was used in order to perform the simulations. In order to evaluate the difference between muscle force curves in the healthy and the drop-foot populations, an integrals of each muscle curve were computed.
Results: The group of patients with drop-foot exhibited an increased force integral for all muscle groups, except for the ankle evertors. The highest increases were observed for hip adductors (112%), hip extensors (88%), knee and hip flexors (83% and 50%, respectively) and for the plantarflexor (47%). These results were mainly influenced by the following muscles: flexor digitorum and hallucius, tibialis posterior and semitendinosus. The force integral for these muscles increased by more than 200% in the drop-foot group as compared to the control group. In addition, significant changes (> 100%) were noted for the posterior thigh muscle group (semitendinosus, biceps femoris long and short head), which are responsible for bending the knee joint and straightening the hip joint.
Conclusions: It was proved that the loss in muscle force in individual muscle groups of the ankle joint are compensated for by the increased force and activity in other muscles acting on this joint and another muscles in neighbouring joints. The results may have important implications for physiotherapy treatments.

1. Introduction

Healthy human gait uses repetitive reciprocal limb motions in order to advance the body while simultaneously maintaining stance stability [1,2]. This is achieved by tightly regulated patterns of muscle activations and generated joint torques and powers. Pathologies that lead to joint deformities, muscle weakness, sensory loss, impaired motor control or pain, interfere with these tightly regulated patterns, and so compensatory strategies might be required in order to maintain proper function. Therefore, if any muscle force development capacity is impaired, it is usually possible to either (1) compensate for the damage to the muscle by modifying the efforts of other muscles, while still performing the same or very similar motion as a whole [3] or (2) significantly modify the motion into another movement that naturally results in a lower mechanical load being imposed on the damaged muscle [4]. In a clinical situation in which a patient chooses the first option [5], it is important to quantitatively evaluate the impact of the damage to a muscle on the other components of the musculoskeletal system, in order to avoid causing additional physical problems [6], and also to learn to what extent the other muscles need to be trained in order to compensate for the role of the damaged muscle.

The impact of muscle deactivation on the biomechanics of the musculoskeletal system can be assessed by the use of inverse dynamics, by which the torques developed around the joints can be calculated using the kinematic motion as an input. The human musculoskeletal system is endowed with a degree of redundancy, meaning that the system has more muscles than degrees of freedom [7], so that an almost infinite number of combinations of muscle forces are possible in order to achieve a given combination of joint torques. In many studies, this redundancy has been developed primarily through optimization

* Corresponding author.
E-mail address: michalinablazkiewicz@gmail.com (M. Błażkiewicz).

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approaches, such as minimizing the level of muscle activation [8], minimizing muscle stress [9] and the pseudo-inverse method [7]. It has been reported that methods based on inverse dynamics yield a highly realistic output for motions such as the human gait [8]. Therefore, recently many researchers have analyzed the contribution of some specific muscles to gait performance [10,11]. Neptune et al. [12] analyzed the role of the plantarflexor muscles during gait, and they calculated the degree to which these muscles contribute to propelling the trunk in the forward direction. In order to determine whether backward walking represents a simple reversal of forward walking, Blazkiewicz [13] computed the muscle force distribution during both types of gait. Piazza and Delp [14] examined the contribution of muscle forces to knee flexion during the swing phase of a normal gait. Moreover, musculoskeletal simulations have been used to explore possible compensatory strategies in response to the reduced force capability of a single muscle. Goldberg and Neptune [15] used a forward dynamic simulation of normal walking to observe muscle compensations in response to the weakness in plantarflexor, quadriceps and hamstrings over a gait cycle. Their study highlighted the ability of the plantarflexor muscle group to compensate for weak hip and knee flexors and extensors, but the model was unable to reproduce a normal walking pattern when the plantarflexor strength was reduced as a group. A study by Jonkers et al. [16] examined the contributions of individual muscles during stance, concluding that a combination of multiple muscles is likely required to compensate for weakness in a single muscle. A study by Steele et al. [17] determined the minimum isometric force requirements for various muscle groups in children with cerebral palsy during crouch gait, before crouch gait becomes impossible.

To our knowledge, no study has yet investigated the effects of dorsiflexor weakness on muscle force distribution and compensations during gait in persons with drop-foot. Drop-foot is a motor deficiency caused by total or partial central paralysis of the muscles innervated by the common peroneal nerve. It is characterized by a lack of voluntary control of ankle dorsiflexion and subtalar eversion, and an abnormal gait pattern [18]. Blazkiewicz et al. [19] undertook to describe the issue of compensation in the drop-foot population, but solely based on temporal-spatial, kinematic and kinetic gait parameters. Therefore, the aim of the present study was to examine the compensatory actions of other muscles in response to dorsiflexion muscle force weakness in persons with drop-foot.

2. Materials and methods

2.1. Participants

The study group (DF) consisted of ten persons with unilateral drop-foot. The mean age was 52.4 ± 13.8, height 177.8 ± 9.4 cm and body mass 74.8 ± 15.7 kg. The subjects suffered from paresis of the common peroneal nerve caused by lumbar disc hernia. In daily life none of the subjects used an ankle joint orthosis to compensate for the loss of dorsiflexion. In contrast, the control group (C) consisted of ten healthy subjects with a mean age of 29.6 ± 9.7, height of 176.3 ± 9.9 cm and weight of 73.5 ± 11.7 kg. All subjects gave their informed written consent to the experimental procedures, which were approved by the local ethics committee.

2.2. Instrumentation and data collection

All subjects underwent a standard three-dimensional gait analysis. Motion data was collected using a 9-camera system (Vicon Motion Systems, UK) operating at 100 Hz, with a standard Plug-in-Gait marker set. Two force plates (Kistler Holding AG, Switzerland) were used to determine ground reaction forces (GRF), which were sampled at 1000 Hz. The force plates were synchronized to the motion capture system. These systems were calibrated according to the manufacturers' recommendations before the trials were conducted. All subjects walked barefoot at their self-selected speed. The analysis was carried out based on the attempts without any random mistakes, with the individual performing the task naturally.

2.3. Dynamic simulations

The representative gait cycle for each subject from the DF and C groups was generated using OpenSim software [20]. A generic musculoskeletal model with 23 degrees of freedom and 92 muscle-tendon actuators [20,21] was used in order to perform simulations. The degrees of freedom in the model included a ball-and-socket joint located at approximately the third lumbar vertebra between the pelvis and torso, ball-and-socket joints at each hip, a planar joint with coupled translations at each knee, and revolute joints at each ankle [22]. The generic model was scaled to individual subject sizes, using the anatomical landmarks and functional joint centers as a reference. The subjects' gait patterns were reproduced by the scaled model, using an inverse kinematic analysis tracking individual marker trajectories. Joint torques were calculated using inverse dynamics. Dynamic inconsistency between measured ground reaction forces (GRF) and kinematics was resolved using a Residual Reduction Algorithm (RRA), by applying small external forces and torques (i.e. residuals) to the torso and making small adjustments to the model's mass properties and kinematics [20]. After running the RRA, the residual forces accounted for less than 10% of the body weight. Computed Muscle Control (CMC) was used to calculate actuator excitations that would track the experimental kinematics. At each time step, CMC determines the set of muscle excitations that produce the forces that generate measured accelerations for all degrees of freedom, taking into account muscle activation dynamics and force-length-velocity properties. The distribution of muscle excitations among redundant actuators was determined by minimizing the sum of activations squared at each time step. Idealized torque actuators, known as reserve actuators, were included for each degree of freedom in the model to provide extra torque if the muscles could not generate the measured accelerations. Reserve actuator torques were used to determine when the muscles could no longer produce the subject's motion. To assess the accuracy of the simulation we compared the simulated kinematics and kinetics to experimental data. The simulated joint angles reproduced the joint angles calculated by inverse kinematics with a maximum difference of less than 3 degrees. Additionally, the results of RRA and CMC were considered acceptable if the model kinematics differed from experimentally measured kinematics by less than 5 degrees and if the peak of residual forces and moments at the pelvis were less than 10–20 N and 75 Nm, respectively [17,20,23]. For all studied persons, the simulation results meet the above criteria.

2.4. Data analysis

To determine the differences between the C and DF groups, and thus to assess the degree of compensation, we analyzed the area under the muscle force curves in the gait cycle domain. The 43 muscles analyzed came from the right lower limb in the healthy individuals, and from the dysfunctional limb in the drop-foot group. All regions were computed in MatLab 2016a (The MathWorks Inc., USA), using numerical integration via the trapezoidal method: $I = \int_{i=1}^{100} |F_i| \, dGaitCycle\,[\%]$, where F – muscle force curve. This method approximates the integration over an interval by breaking the area down into trapezoids with more easily computable areas. The results obtained for the two populations were compared using the non-parametric U Mann-Whitney tests. All statistical analysis was performed using Statistica 12 (StatSoft. Inc., USA), with the significant p-value set at 0.05.

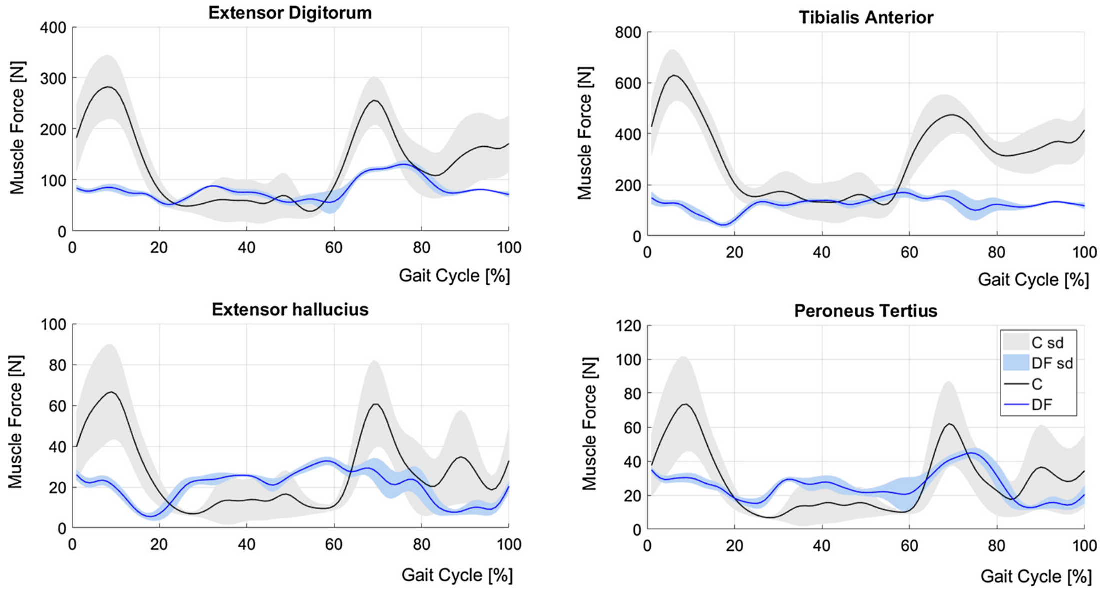

Fig. 1. The average and standard deviation shapes of the time-sequence muscle force curves in Drop-Foot (DF) and Control (C) group of 4 dorsiflexor muscles: tibialis anterior, extensor digitorum, extensor hallucius and peroneus tertius.

3. Results

3.1. Weakened dorsiflexor muscles

Each of the four weak individual dorsiflexor muscles in the DF group produced almost half the force in comparison to the C group (Fig. 1).

Based on the value of the integral of the force curves of dorsiflexor muscles, we found statistically significant differences between groups C and DF ($p = 0.0021$) in terms of muscular forces: tibialis anterior, extensor digitorium and extensor hallucius (Fig. 2A). In the control group for these muscles, values of force integrals were noted to be larger by 148%, 60% and 37% respectively, calculated based on the gait cycle. The peroneus tertius muscle had only about 15% higher values compared to those achieved in the DF group. However, the main differences between the groups occurred at (0–20)% of the gait cycle (GC) and during the swing phase (Fig. 2C). The largest differences were found for muscle tibialis anterior (414% during loading response phase and 194% in swing phase) and for muscle extensor hallucius (202% and 72%). In contrast, in the midstance and terminal stance (20–60)%GC muscles produced similarly low strength levels in both the drop-foot group and healthy individuals. The average value of the percentage difference in this phase for four muscles is 56%.

Looking at the individual muscle groups affecting the ankle joint (Fig. 2B), statistically significant differences between the group of healthy and disabled individuals were observed for dorsiflexors ($p = 0.0021$), plantarflexors ($p = 0.0072$) and ankle invertors ($p = 0.0214$). In healthy subjects, the value of integral of muscles force in dorsiflexors and evertors was respectively 98% and 25% greater than in the DF group in gait cycle. As before, the loading response phase mainly affect the values of these differences (Fig. 2D), by 270% and 24% respectively.

In response to the dysfunction, individuals with drop-foot generated 47% and 25% greater integral value for plantarflexors and invertors in all gait cycle, respectively, as compared to the C group. These results were influenced mainly by the following muscle groups: flexor digitorum, flexor hallucius and tibialis posterior. For these muscles, the value of force integral in the DF group was larger by 288% (289%, 676% and 106%), 280% (307%, 557%, 67%), and 208% (199%, 326%, 46%), respectively, compared to the control group in all gait cycle ((0–20)%GC, (20–60)%GC, swing phase). In addition, statistically significant differences at $p = 0.0032$ were also noted for all three muscles.

3.2. Compensatory muscles

In response to the dorsiflexor weakness, greater integral values of muscles force were noted for all muscle groups in the knee and hip joints in the DF group (Fig. 3A, B). While statistically significant differences between groups C and DF were observed for knee flexors ($p = 0.0021$), hip flexors ($p = 0.0106$), hip extensors ($p = 0.0049$) and hip adductors ($p = 0.0021$).

In the DF group, the value of the integral of muscle forces of the hip adductors and extensors group was 112% and 88% greater than in the C group, respectively, and for the knee and hip flexors it was larger by 83% and 50%, respectively (Fig. 3A).

Such results are influenced by the fact that the weakening of 4 dorsiflexors caused all the remaining 39 muscles in the lower extremity model to change their force output. As many as 32 muscles in the DF group were seen to increase the value of the muscles force integral in the gait cycle. Among them, there were statistically significant differences ($p < 0.05$) for 20 individual units. The greatest differences ($p = 0.0021$) were calculated for 6 double joint muscles in the knee and hip joints (Fig. 4), belonging to the posterior thigh group.

Four muscles acting as both knee flexors and hip extensors increased the force integral value in the DF group by 270% in gait cycle (50% in (0–20)%GC, 97% in (20–60)%GC, 213% in swing phase) – semitendinosus, 134% in GC (249%, 96%, 164% in each phase, respectively) and 127% in GC (14%, 5%, 131% in each phase, respectively) – biceps femoris long and short head, and 79% in GC (173%, 26%, 69% in each phase, respectively) – semimembranosus. The remaining two double joint muscles – gracilis and sartorius, knee and hip flexors – increased the force integral value by 167% and 63%, respectively in GC (Fig. 4). The remaining muscles that significantly affected the results were mainly hip joint flexor muscles: adductor longus and brevis (117%), psoas, iliacus and tensor fasciae latae (80%) in GC.

4. Discussion

The purpose of this study was to evaluate changes in muscle forces distribution during the gait cycle in response to dorsiflexor weakness in patients with drop-foot. To our knowledge, this is the first time muscle-

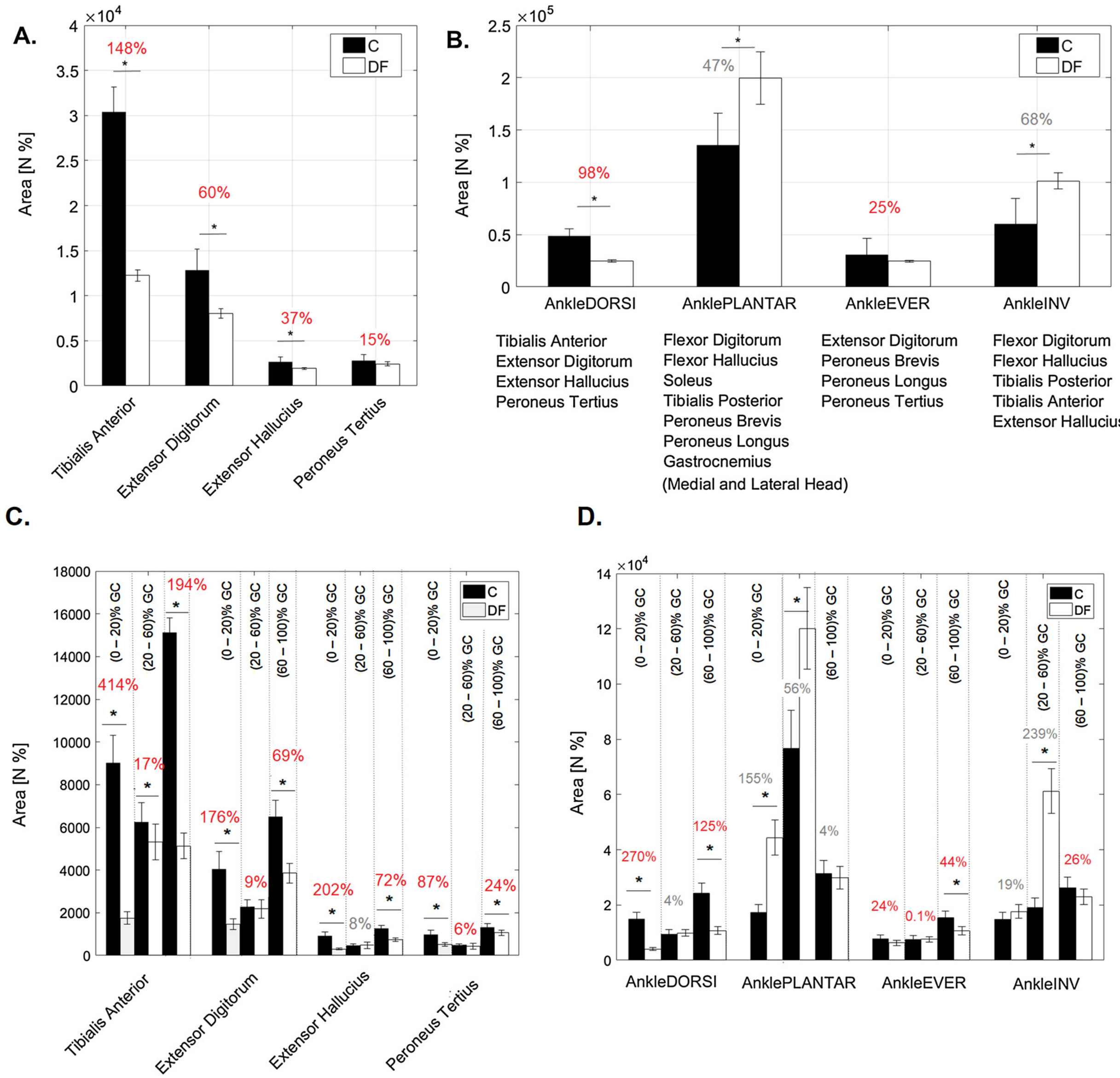

Fig. 2. The values of the force integrals in the control group and the drop-foot group for: A. four dorsal foot flexors, B. the dorsiflexors (AnkleDORSI), plantarflexos (AnklePLANTAR), evertors (AnkleEVER), invertors (AnkleINV) groups, and the list of their muscles, C. four dorsiflexors in the phases of the gait cycle, D. all ankle muscle groups in the phases of the gait cycle. Percentages are the differences between groups in: all gait cycle (A., B.) and in individual phases of the gait cycle (C., D.); * - statistically significant differences ($p < 0.05$).

driven simulations were used to investigate how lower extremity muscles would work in persons with drop-foot, as compared to healthy individuals. The muscle force curves in our simulations for healthy persons compared well with the muscle forces in previous studies [12,21,23,24].

There are several reports in literature that deal with the issues of evaluation of the influence of muscle deactivation on other muscles and joints during gait. Most works assess how much weakness could be tolerated before execution of normal gait became impossible. van der Krogt et al. [23] revealed that normal walking is remarkably robust to the weakness of some muscles, yet sensitive to others. Weakness of hip and knee extensors can be well tolerated without an extensive increase in muscle stress, but in contrast, gait is most sensitive to weakness of plantarflexors, hip flexors and abductors. Moreover, when all muscles were weakened simultaneously, they found that activation increased in all muscles. But, when individual muscles were weakened, these muscles generally produced less force and other muscles compensated by increasing their force. Similar results were obtained in our work. In our DF group, as a result of weakening the dorsiflexor muscles, the value of the generated force changed for all muscles in comparison to the healthy group. As many as 32 of the 39 muscles in the drop-foot group increased value of the muscles force integral in the gait cycle, with statistically significant intergroup differences ($p < 0.05$) being observed for 20 muscular units.

It is worth noting that in their study, van der Krogt et al. [23] weakened only one anterior tibialis muscle from the dorsiflexor group, which resulted in the extensor digitorum and peroneus tertius muscles increasing their activation. This seems plausible because these muscles work within one group and are closest to the anterior tibialis muscle. Knarr et al. [25] used musculoskeletal simulations to explore compensatory strategies in a post-stroke group. In this population, during the swing phase of gait, drop-foot is a common issue associated with dorsiflexor weakness [26]. Simulations of three dimensional dynamics were created from 10 healthy subjects and constraints were set on the activation capacity of the plantarflexor, dorsiflexor and hamstrings muscle groups to simulate activation impairments. The model was

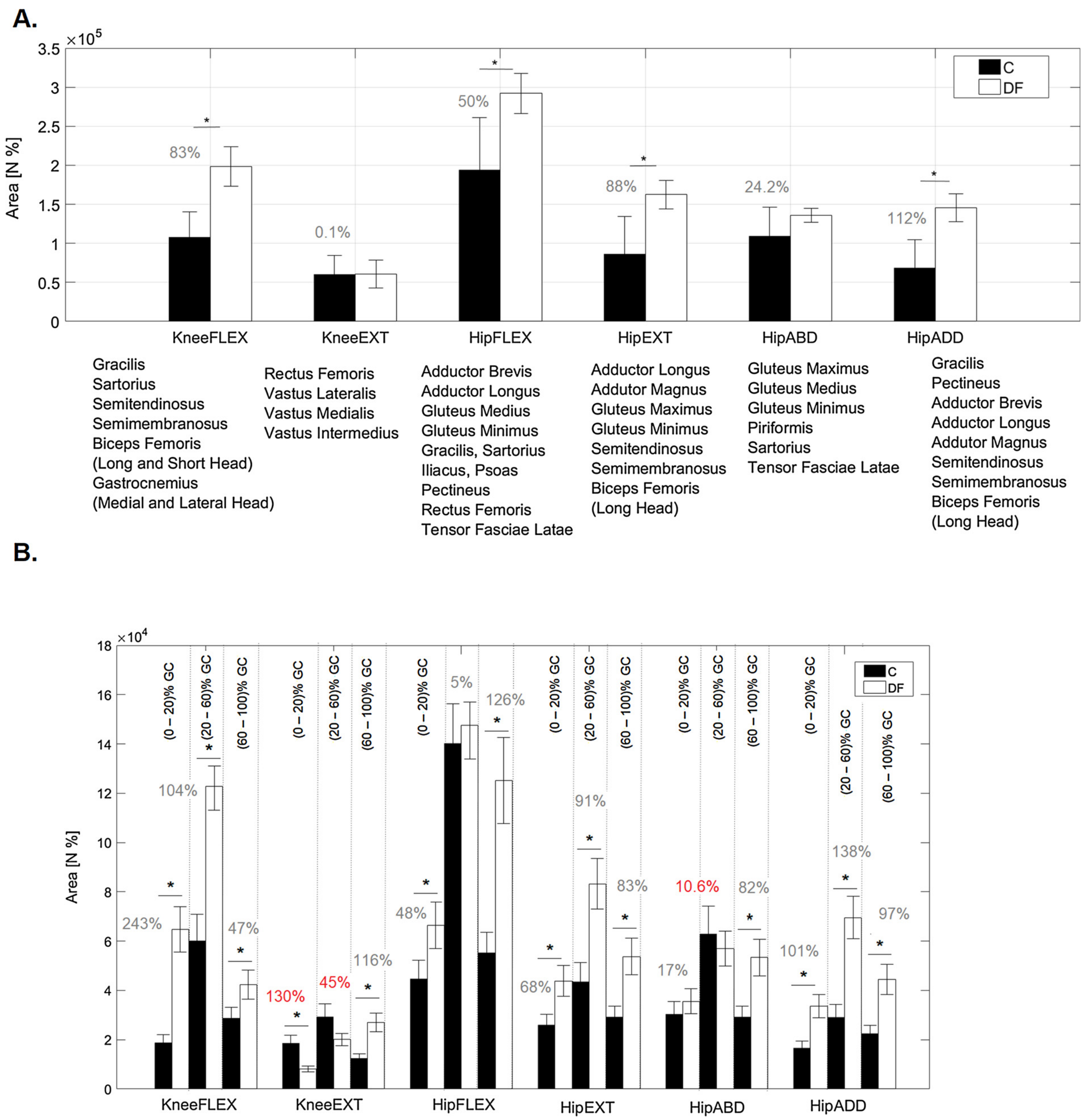

Fig. 3. The integral values of muscle forces affecting the knee and hip joint in the group of healthy individuals and persons with drop-foot, A. in the gait cycle and the list of muscles involved, B. in loading response phase (0–20)%GC, mid and terminal stance (20–60)%GC and swing phase (60–100)%GC. Percentages are the differences between groups; * - statistically significant differences ($p < 0.05$).

unable to recreate the normal gait pattern with simultaneous impairment of all three muscle groups. But when the muscle groups were impaired individually, the model required that the plantarflexor, dorsiflexor and hamstrings muscle groups were activated to at least 55%, 64%, and 18%, respectively, to recreate the subjects' normal gait pattern. Four muscles (biceps femoris short head, tensor fasciae latae, pectineus, rectus femoris) showed a change of more than 5% in percent activation between the normal and dorsiflexor impaired model during the first half of swing. The biceps femoris short showed the largest increase in activation at 19.5 ± 4.3%, followed by the tensor fasciae latae at 13.3 ± 8.8%. In our study, dorsiflexor weakness increased the value of the muscle force integral in the plantaflexors group by 288% for flexor digitorum, 280% (flexor hallucinus) and 208% (tibialis posterior) in comparison with the control group. There were no results in any of the cited models that would emphasize such large changes in the generated force for the plantarflexor group. On the other hand, each of these models simulated such muscle weakness or muscle groups that still allowed the kinematics of proper walking to be maintained. In our study, the dorsiflexor weakness was severe and patients choose to alter their own gait pattern [19] rather than to walk with normal kinematics using the compensations as found in studies [23,25]. Therefore, compensations for weakness were seen in many different muscles and muscle groups throughout the leg.

Following large increases of force for the plantarflexor group, significant changes in the values of the generated forces were observed, primarily for the double joint muscles. In the DF group, the four posterior thigh muscles serving as knee and hip flexors increased the value of the force integral by 270% (semintendinsous), 134% and 127% (biceps femoris long and short head), and 79% (semimembranosus). In contrast, the remaining two double joint muscles, gracilis and sartorius, the knee and hip flexors, increased the value of the muscle force integral by 167% and 63%, respectively. The remaining muscles with

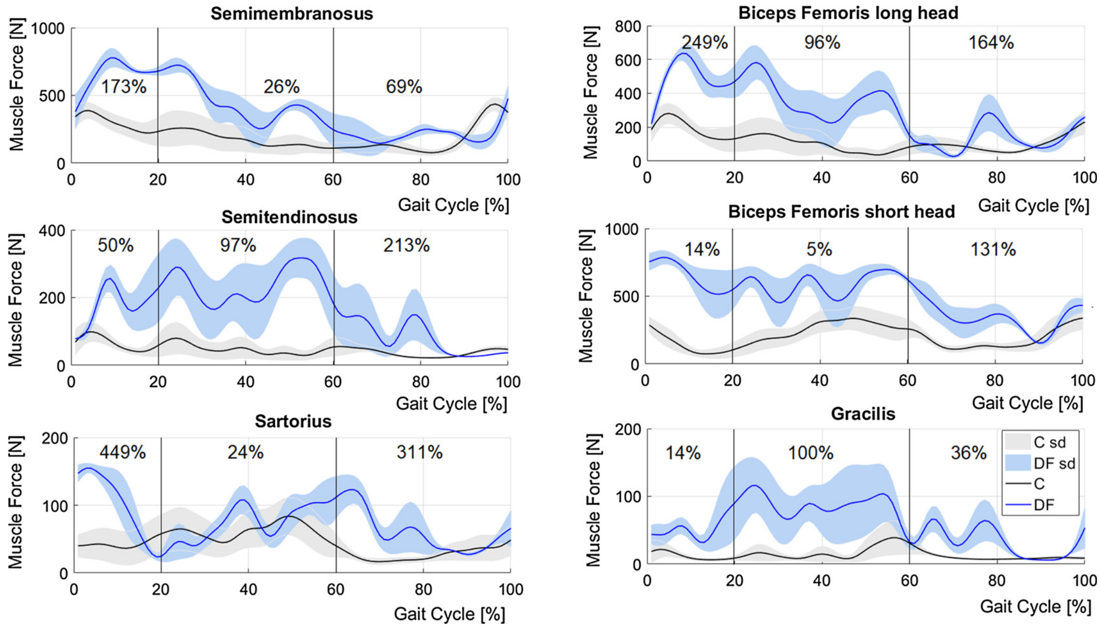

Fig. 4. The average and standard deviation muscle force shapes in Drop-Foot (DF) and Control (C) group of 6 biarticular muscles: semitendinosus, semimembranosus, sartorius, biceps femoris long and short head, gracilis. Percentages are the differences between groups.

Table 1
Overview of the work of muscles and muscle groups in the DF group where: ↓, ↑ represents a decrease or increase in force in the Drop-Foot group compared with the Control group in gait cycle domain. Percentages are the differences between groups in individual phases of the gait cycle, according to the scheme: ((0–20)%GC, (20–60)%GC, swing phase).

Muscle Group		Joint		Individual Muscles
Dorsiflexors Evertors	↓ (98%) ↓ (25%)	Ankle	↓	Tibialis Anterior (63%, 17%, 195%), Extensor Digitorum (202%, 19%, 69%), Extensor Hallucius (202%, 102%, 72%), Peroneus Longus (87%, 31%, 36%), Peroneus Tertius (176%, 87%, 24%), Soleus (87%, 87%, 124%).
Plantarflexors Invertors	↑ (47%) ↑ (68%)		↑	Peroneus Brevis (48%, 0.5%, 17%), Flexor Digitorum (202%, 102%, 172%), Flexor Hallucius (414%, 557%, 66%), Tibialis Posterior (176%, 19%, 169%), Gastrocnemius Medial Head (63%, 0.5%, 18%), Gastrocnemius Lateral Head (48%, 31%, 36%).
Flexors	↑ (83%)	Knee	↓	Vastus Medialis (289%, 676%, 106%), Vastus Intermedius (307%, 557%, 67%).
Extensors	↓ (0.1%)		↑	Gracilis (14%, 100%, 36%), Sartorius (449%, 24%, 311%), Semitendinosus (50%, 97%, 213%), Semimebranosus (173%, 26%, 69%), Biceps Femoris Long Head (249%, 96%, 164%), Biceps Femoris Short Head (14%, 5%, 131%), Rectus Femoris (199%, 326%, 46%), Vastus Lateralis (414%, 18%, 195%).
Flexors	↑ (50%)	Hip	↓	Piriformis (19%, 35%, 105%).
Extensors Abductors Adductors	↑ (88%) ↑ (24.2%) ↑ (112%)		↑	Adductor Brevis (28%, 2%, 30%), Addutor Magnus (120%, 32%, 57%), Adductor Longus (337%, 399%, 210%), Gluteus Minimus (58%, 52%, 153%), Gluteus Medius (239%, 270%, 84%), Gluteus Maximus (142%, 113%, 95%), Tensor Fasciae Latae (156%, 22%, 122%), Iliacus (338%, 21%, 59%), Psoas (340%, 11%, 82%), Pectineus (57%, 19%, 38%).

significant results were mainly hip joint muscles: adductor longus and brevis (117%), psoas, iliacus and tensor fasciae latae (80%). The present study demonstrated that for all muscles, the loading response phase (0–20)%GC and swing phase is the key phase. In this phases, the large differences in values of integrals between both groups for all muscles are visible. For weakened muscles, the differences range from 87% to 414% for (0–20)%GC and 24%–194% in swing phase (Fig. 2C). While, for two double joint muscles, the differences range from 14% to 449% for loading response phase and 36% to 311% (Fig. 4). The work of muscles and muscle groups observed in this study, summarized in Table 1, is supported by the gait kinematics of the drop-foot population, where early swing is a contributing factor to the selection of abnormal gait patterns such as hip hiking and circumduction, which are commonly used to accommodate foot drop by increasing swing phase foot clearance. Therefore, the results presented in our study indicate that in the drop-foot group kinematic compensations become more favorable than force generation compensations in maintaining normal kinematics.

5. Conclusions

In summary, in the drop-foot population, changes in generated force are seen in the muscle force across multiple muscle groups. In this paper we proved that the loss in muscle force in individual muscle groups of the ankle joint are compensated for by the increased force and activity in other muscles acting on this joint, i.e. the tibialis posterior, flexor hallucis and digitorum, and another muscles in neighbouring joints i.e. semitendinosus and semimembranosus. Therefore, our results may have important implications for developing physiotherapy treatments.

Conflict of interests

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

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