Modelling of human walking to optimise the function of ankle-foot orthosis in Guillan-Barré patients with drop foot

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ABSTRACT

Introduction: This paper deals with the dynamic modelling of human walking. The main focus of this research was to optimise the function of the orthosis in patients with neuropathic feet, based on the kinematics data from different categories of neuropathic patients.

Methods: The patient's body on the sagittal plane was modelled for calculating the torques generated in joints. The kinematics data required for mathematical modelling of the patients were obtained from the films of patients captured by high speed camera, and then the films were analysed through a motion analysis software. An inverse dynamic model was used for estimating the spring coefficient.

Results: In our dynamic model, the role of muscles was substituted by adding a spring-damper between the shank and ankle that could compensate for their weakness by designing ankle-foot orthoses based on the kinematics data obtained from the patients. The torque generated in the ankle was varied by changing the spring constant. Therefore, it was possible to decrease the torque generated in muscles which could lead to the design of more comfortable and efficient orthoses.

Conclusion: In this research, unlike previous research activities, instead of studying the abnormal gait or modelling the ankle-foot orthosis separately, the function of the ankle-foot orthosis on the abnormal gait has been quantitatively improved through a correction of the torque.

Keywords: ankle-foot orthosis, gait analysis, Guillan-Barré syndrome, steppage gait

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Table I. Estimate of the mass, length, centre of mass and moment of inertia of the five body segments under study.

Body segment	Mass (kg)	Length (m)	Centre of mass (m)	Moment of inertia (kg.m^2)
Toe	0.569	0.0588	0.0353	$1.851e^{-5}$
Ankle	3.123	0.1763	0.1058	$8.729e^{-4}$
Shank	4.475	0.4113	0.2414	0.0068
Thigh	15.961	0.5287	0.3183	0.04016
Trunk	17.761	0.7050	0.2965	0.79448

INTRODUCTION

In recent years, considerable attention has been paid to studying human walking. In particular, there has been increasing enthusiasm to research human walking due to the predictive capabilities and potential applications in the areas of clinical biomechanics, rehabilitation engineering, neurosciences and robotics.[1-8] A very important aspect of gait analysis is the investigation of the kinetics of human body segments from kinematics data or vice versa. Walking involves a pendular exchange between potential energy and kinetic energy, resembling the action of an inverted pendulum. This seemingly simple task exhibits complex characteristics. During walking, the human body is characterised by periodic dynamics involving mechanical energy exchanges, and its natural frequency is an important determinant of spontaneous walking.[9] Normal walking depends on a continual interchange between mobility and stability. Free passive mobility and appropriate muscle action are basic constituents. Any abnormality restricting the normal free mobility of a joint or altering either the timing or intensity of muscle action creates an abnormal gait. An abnormal gait may be due to an injury, disease, pain or problems of motor control.[10] The ability of the orthosis to compensate for the abnormality determines the amount of functionality retained.

Drop foot is an abnormal neuromuscular disorder that is characterised by a steppage gait that affects the patient's ability to raise their foot at the ankle, and is further

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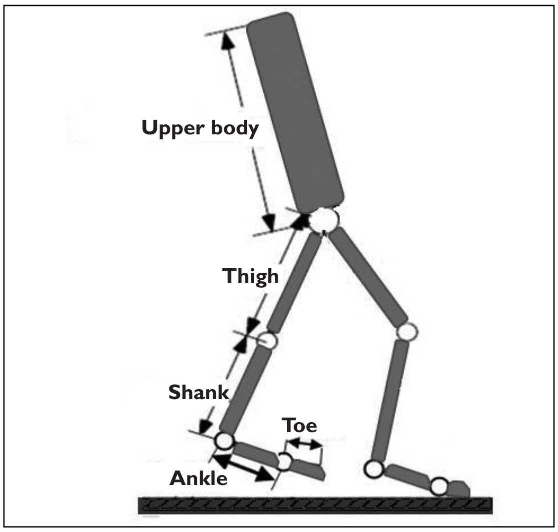

Fig. 1 Diagram shows the nine segments of the human body.

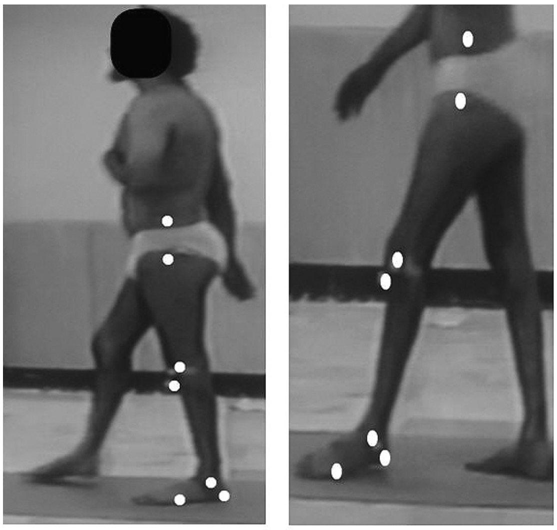

Fig. 2 Photographs show the marker locations on a male patient's joints.

characterised by an inability to point the toes towards the body (dorsiflexion) or move the foot at the ankle inwards or outwards.[11] The causes of foot drop, going from damage to the peripheral to central nerves, include: muscle disease, peroneal nerve (common), sciatic nerve, lumbosacral plexus, L5 nerve root, spinal cord (rarely causes isolated foot drop), brain (uncommon, but often overlooked), and non-organic.[12] An example of a peripheral nerve disorder is Guillan-Barré syndrome (GBS), which is a heterogeneous grouping of immune-mediated processes generally characterised by motor, sensory and autonomic dysfunction. In its classic form, GBS is an acute inflammatory demyelinating polyneuropathy characterised by progressive symmetric ascending muscle weakness, paralysis and hyporeflexia with or without sensory or autonomic symptoms. However, variants involving the cranial nerves or pure motor involvement are not uncommon. In severe cases, muscle weakness may lead to respiratory failure and labile autonomic dysfunction.[13]

Drop foot patients can be fitted with an ankle-foot orthosis, brace or splint to stabilise the ankle-foot. An ankle-foot orthosis is a device that is applied externally to the calf band of the patient to improve function of the musculoneuroskeletal system. Previous research has indicated that the design of the ankle-foot orthosis was based on a sample prototype, and evaluation was in accordance to the kinetic and kinematic effects during the walking.[14-22] The purpose of this study was to quantitatively optimise the function of the orthosis on steppage gait based on the inverse kinematics method. Unlike previous research, instead of studying the abnormal gait or modelling the ankle-foot orthosis separately, the function of the ankle-foot orthosis on the abnormal gait was quantitatively studied. In this approach, a simulation of the human gait was generated by combining the orthosis function and steppage gait. Finally, the optimum parameters of ankle-foot orthosis was calculated.

METHODS

For mechanical modelling of the body, various dynamic models with different inputs and outputs for the simulation of human movement have been developed. Since internal forces and torques in joints could not be measured directly in a biological system, the kinematics and anthropometric parameters were used because they were calculated indirectly. The mass of the body segments was determined through the Zatsirosky approach.[23] In order to estimate the length of each segment, the formulations by Muftic and Seif were used.[24] In order to estimate the moment of inertia around the centre of the mass, the Chaffin and Andersson formula[25] was used:

$$I = 0.09ML^2$$

where M is the mass of the segment and L is the length of the segment. The patient was male and his height, weight and age were 188 cm, 66.215 kg and 18 years, respectively. The results of estimating the mass, height and centre of the mass segment are shown in Table I. The 60 gait cycles of the patient were used to reach a correct statistical estimation. The patient walked at self-selected normal velocities with no orthosis.

For momentum calculation from a dynamic model, a sagittal model of nine segments of freedom of an anthropomorphic biped was used (Fig. 1). In this research, the body in the sagittal plane was divided into five segments: toe, ankle, shank, thigh and upper body. The body's mass, moment of inertia tensor of the patient with

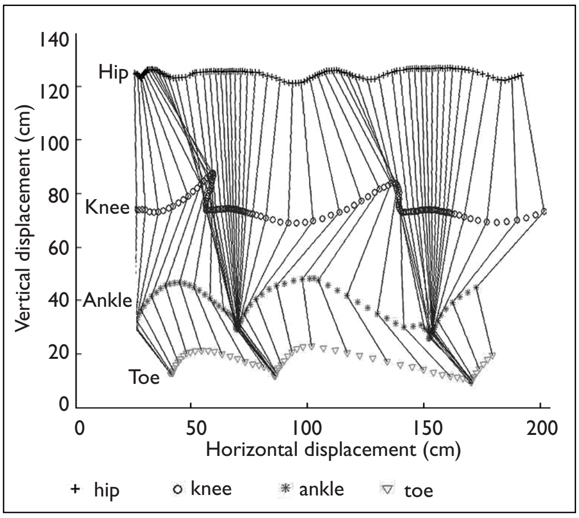

Fig. 3 Graph shows the motion of the patient during a complete gait cycle captured by the motion analysis system.

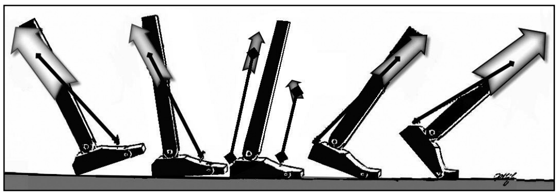

Fig. 4 Diagram shows the effect of ankle motion, controlled by muscle action, on the pathway of the knee.[28]

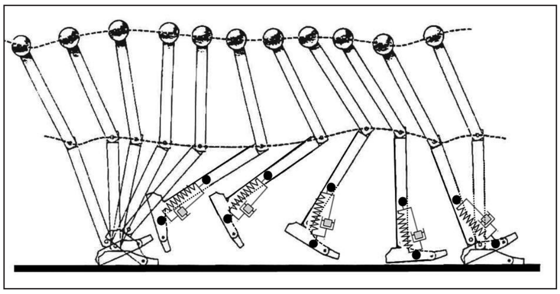

Fig. 5 Diagram shows the schematic of the gait cycle for optimising the function of the ankle-foot orthosis.[29]

Fig. 6 Photograph shows a sample of ankle-foot orthoses.[30]

respect to the body's centre of mass and kinematics data were required for the modelling and simulation of each segment in the dynamic model using a SimMechanics toolbox.[26] By placing sensors on the SimMechanics model and using the following equations, the desirable outputs, including matrices of torque in the joints, were calculated during continuous modelling. It was assumed that the segments were rigid links, the length of each segment was constant during the simulation, the friction in joints ignored, and every joint considered as a revolute joint. The required kinematics data that was obtained from the film of the patient on the sagittal plane were captured by a high-speed camera that set the frame rate at 50 Hz. Finally, the recorded data was analysed by a motion analysis software.[27]

The male patient wore shorts and no shoes. Retroreflective markers were placed on the skin over the head of the fifth metatarsal, the ankle (distal lateral malleolus), the knee at the lateral femoral condyle, the greater trochanter of the hip, and the mid-superior iliac crest directly superior to the greater trochanter of the hip (Fig. 2). A reflective ball (2 cm in diameter) was placed on the walkway to mark the x-axis of the force platform. As shown in Fig. 2, the patient needed to raise one hand and place his arm near his chest so that the camera could capture all the markers.

The kinematics data obtained is presented in Fig. 3. The dynamic model was solved using an inverse kinematics method. In fact, provision of the normal ankle, as shown in Fig. 4, with proper phasing of the extensor and flexor muscles and only a minor amount of motion occurring in the ankle joint, resulted in achievement of a normal pathway of the knee joint.[28] Consequently, patients with drop foot who suffer from muscle disorders do not have a normal pathway of the knee joints and our goal was to substitute the role of the muscles by adding a spring-damper to our model that compensated for their weakness through designing ankle-foot orthosis based on the kinematics data from a GBS patient. By changing the spring-damper coefficients, the torque in the ankle joints of the patients could be approximately similar to the torque in the normal gait. In order to find the best coefficients, the neural network methods were used. The schematic of our model is shown in Fig. 5. In Fig. 6, a sample of ankle-foot orthoses is shown.

A five-link model on the sagittal plane was designed for the simulation of human gait by SimMechanics (Fig. 7). In this figure, the body spring-damper block models

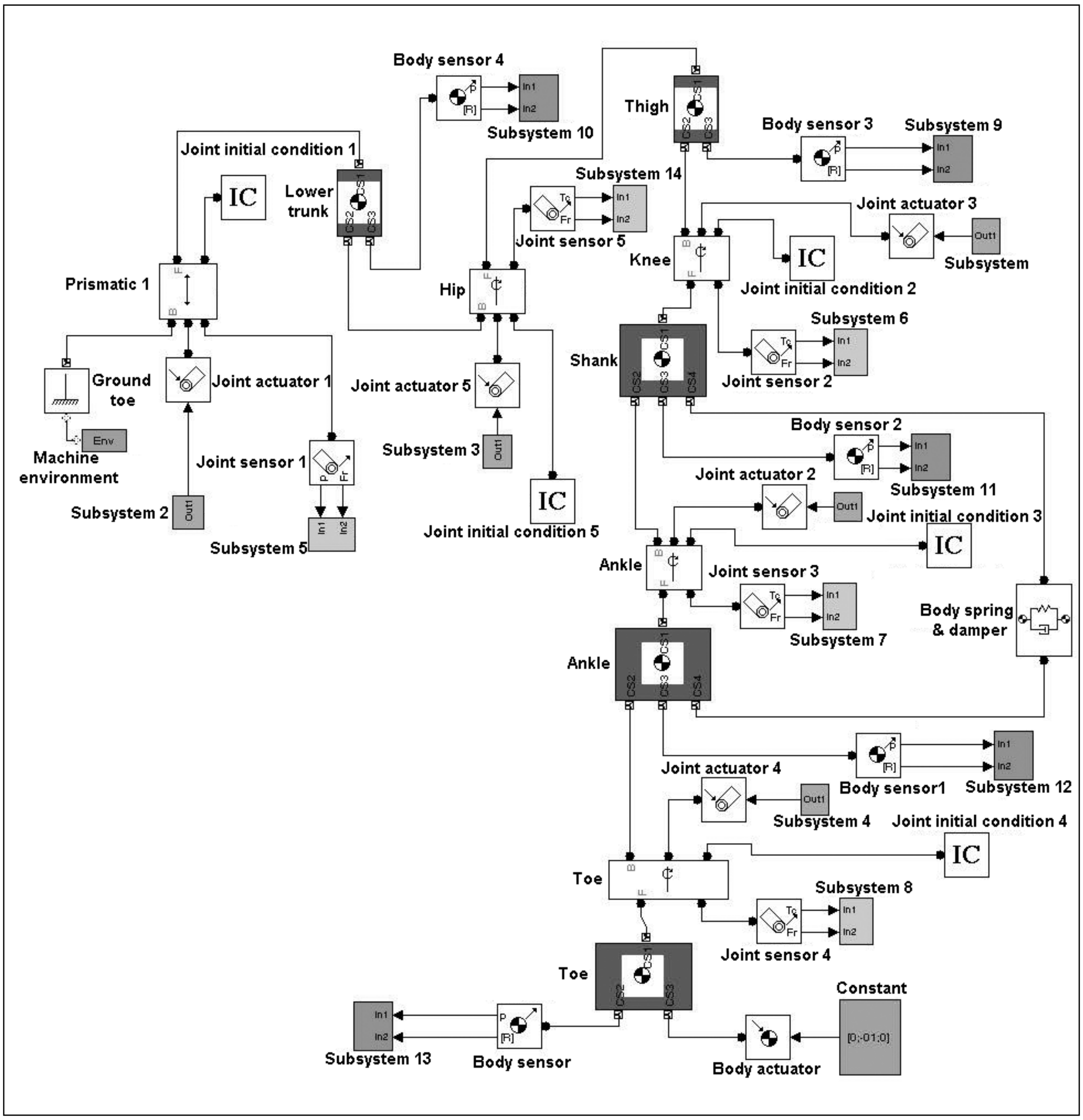

Fig. 7 Chart shows a five-link model on the sagittal plane, designed for a patient by SimMechanics.

the force of the damped spring acting between the two bodies. This block is connected to each body, A or B, at a body coordinate system. If r_A and r_B are the positions of these body coordinate systems, the relative position vector connecting them is $r = r_B - r_A$. The distance of separation is $|r|$. The relative velocity is $v = dr/dt$. Then, the vector force that body A exerts on body B is as shown:

$$F = -k(|r| - r_0)(r / |r|) - b(v \cdot r)(r / |r|^2)$$

where k is the spring constant and b is the damping constant.[26]

RESULTS

In this study, the function of the orthosis on the steppage gait was quantitatively optimised, based on the inverse kinematics method. The simulation of the human gait was conducted by combining the orthosis function and steppage gait. Finally, the optimum parameters of the ankle-foot orthosis were calculated before it was made physically. Fig. 8 illustrates that the computed torques in the hip and knee joints derived from the model were similar to previous works,[31] but the computed torque in the ankle was different because of the neuromuscular disorder in the patient. The torque was captured from the patient when the spring constant was equal to one and the damping constant was equal to zero. As shown in Fig. 9, the required torque for the human gait was decreasing when the spring constant was increasing. In this way, the best value of the spring could be achieved in order to optimise the human gait.

DISCUSSION

In this study, unlike previous research, the function of the orthosis on the abnormal gait was quantitatively

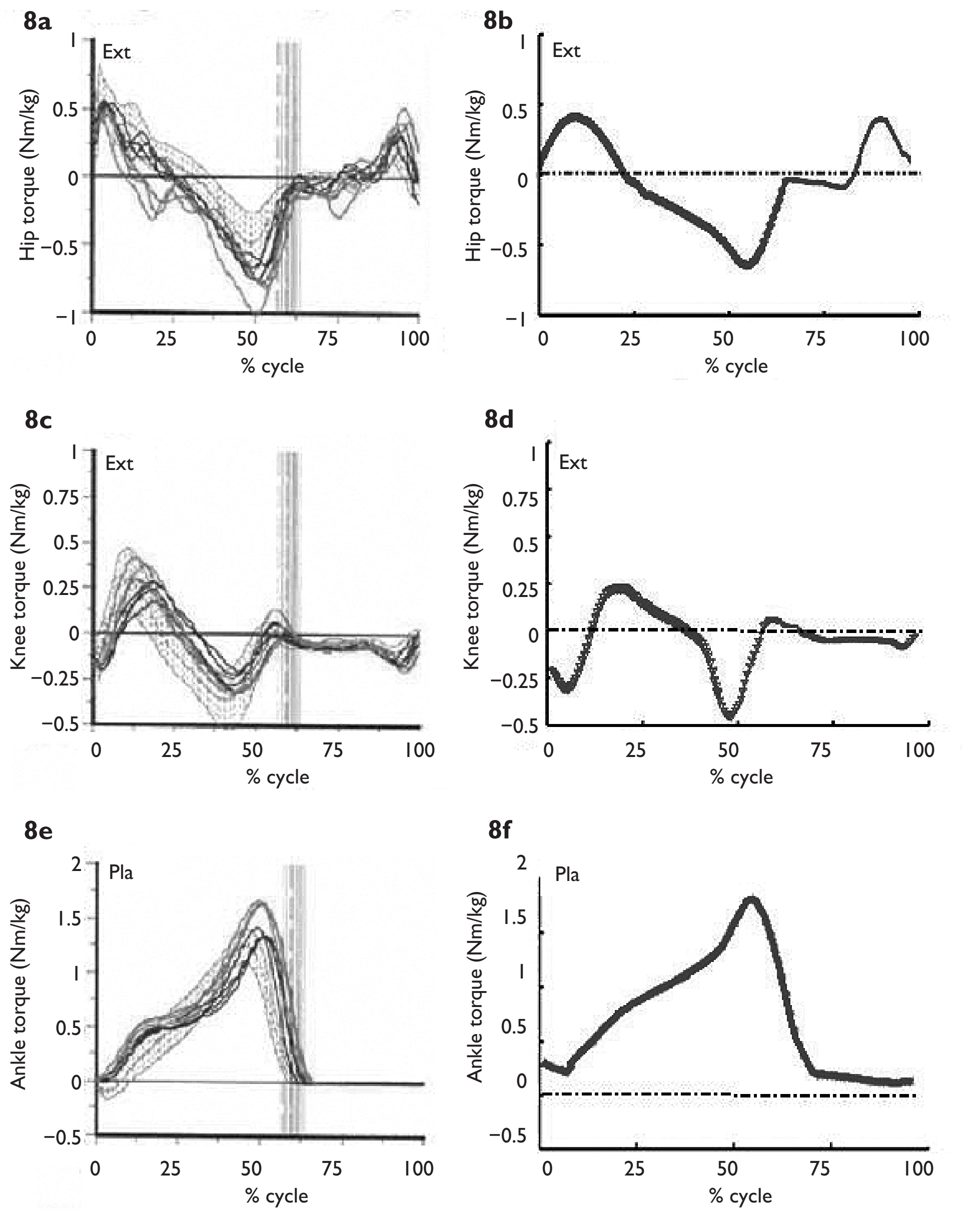

Fig. 8 Line graphs show: (a) Hip sagittal torque from previous research.[31] (b) Hip sagittal torque calculated in our model. (c) Knee sagittal torque from previous research.[31] (d) Knee sagittal torque calculated in our model. (e) Ankle sagittal torque from previous research.[31] (f) Ankle sagittal torque calculated in our model.

calculated. By changing the spring coefficient, it was possible to adjust and decrease the value of the torques generated in the joints. Therefore, the orthosis could be customised for each patient. The goal of this research was not gait quality assessment in GBS patients using a database, it was to develop a human walking model over a complete gait cycle to optimise the function of ankle-foot orthosis in GBS patients with drop foot. The ability to generalise the findings of this research requires further research with more GBS patients. The difference in the patient gait diagram during walking with a raised hand was ignored because of our limitations. However, this difference did not affect the results, because our torque diagram was similar to that of previous research. The connection point of the spring and damper was placed in the middle of the shank and ankle joint. The changing of the connection point may lead us to different functions of the ankle-foot orthosis. It is recommended to assess this point, in order to obtain the precise location of the spring-damper connection to the joints. Future work would involve extending the model for adjusting the damper coefficient in order to reduce the impact generated in the joints. Also, the model can be extended to three dimensions for a precise estimation of the parameters. It is recommended to replicate this research by substituting the linear spring with a non-linear spring.

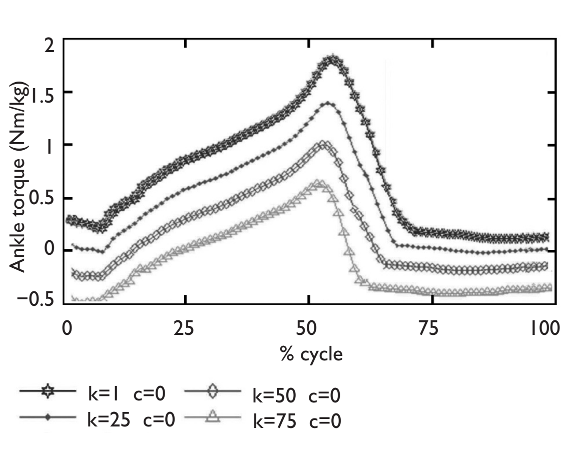

Fig. 9 Line graph shows the effect of changing the spring constant on a calculated torque in ankle joints.

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