Analysis of Hemiparetic Gait by Using Mechanical Models

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Abstract

Clinical gait analysis is applied to analyse the disturbed gait pattern of a patient and to define a suitable therapy. Up to now, mostly measured kinematic gait parameters are considered. The usefulness of kinetic quantities has been recognised, but only scarcely applied. This paper presents a method, derived from robotics, to calculate kinetic quantities of human gait without using expensive force platforms. Kinetic quantities help to better analyse the disturbed gait pattern, to refine diagnosis and assess outcome of therapy. The application of this method in a neurological clinic for analysis of hemiparetic gait is shown.

1. Introduction

Hemiparesis with violent impairment of motricity of one side of the body occurs after stroke or skull-brain trauma. At the beginning and during rehabilitation hemiparetic patients undergo analysis of their gait. In clinical gait analysis the pathological state of the disturbed gait pattern is evaluated first qualitatively and quantitatively, in order to obtain a diagnosis and to determine a suitable therapy. Up to now mostly temporal variables such as stride length, gait speed, walking symmetry [1], kinematic quantities such as joint angles [2] and electromyograms [3] were considered. Despite a large amount of studies, the analysis of kinetic quantities like joint torques, power, work and ground reaction forces was considered only in few studies [4]. The reason therefore resides in the great effort to determine kinetic quantities, because they cannot be directly measured but have to be calculated by inverse models using expensive force platforms. Kinetic quantities, however, enable a further refinement of diagnosis and a better understanding of pathological gait. [5] point out, that kinetic quantities significantly help to separate primary abnormalities (caused by the neurological deficit) from secondary abnormalities (adaptations to circumvent the primary). [6] emphasises, that a multifactorial gait analysis, where kinematic, kinetic and myographic quantities are analysed should be aspired. Furthermore investigations about progression of rehabilitation and specific therapies are missing [4], [7]. [8] describes a method in robotics, which calculates and optimises kinetic quantities when the trajectory of the robot is given.

2. Mechanical Model

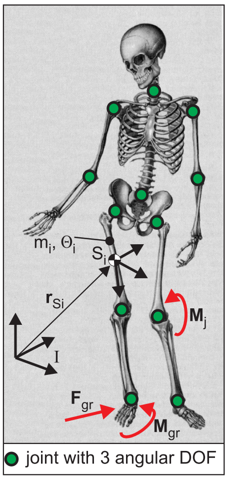

Figure 1: Mechanical model

The human locomotor apparatus is modelled as a three-dimensional system of multiple rigid bodies, connected by ideal ball-and-socket links. Segmentation of the human body respects the anatomy of the locomotor apparatus and the characteristics of human walking. Figure 1 shows the mechanical model, consisting of 13 segments and 12 links, each with 3 degrees of freedom (DOF). With 6 DOF for the trunk, defining its position and orientation in space, the model comprises 42 DOF. The joints are assumed frictionless. A coordinate frame is assigned to each segment, the x axis pointing in the anatomical (upright standing) position forward, the y axis upward and the z axis laterally. In the anatomical position, the coordinate axes of the segments are parallel. The sequence of rotation of each joint is internal/external rotation - adduction/abduction - flexion/extension [9]. The origin of the coordinate system lies in the center of mass of each segment.

The segments are described by their mass, moment of inertia with respect to the body fixed coordinate axes, the vector to the center of gravity and the vector to the distal link. Muscles are not taken into consideration, therefore the mechanical model of the human locomotor apparatus is just a skeletal one.

Several external forces and torques act on the human body when walking, see figure 1. Ground reaction forces $\boldsymbol{F}_{gr}$ and torques $\boldsymbol{M}_{gr}$, which result from the interaction of the foot with the floor, act on the foot in contact. Joint torques $\boldsymbol{M}_j$ summarise all the torques which are generated by the joint spanning muscles.

The mass and inertia properties of the legs and arms are calculated by approximating their shapes with geometric bodies and using average body densities, as described by [10]. By this means, the thigh, shank, upper and underarm are approximated by frustums with circular sections. For the trunk, pelvis, feet and head the regression equations of [11] and [12] are used. By this means the general model of the locomotor apparatus is adapted to each patient. The use of this technique is obligatory because parameters of subjects vary within a wide range, e.g. hemiplegics after stroke who suffer from heavy reductions of the muscle mass in contrast to healthy well-trained persons.

3. Mathematical Formulation

The dynamics of the locomotor apparatus is described in configuration space by the equations of motion. The Newton-Euler method, see [13], with the reference points in the centers of gravity of the segments yields

$$\boldsymbol{M}(\boldsymbol{q},t)\,\ddot{\boldsymbol{q}}(t) - \boldsymbol{h}(\boldsymbol{q},\dot{\boldsymbol{q}},t) = \sum_{i=1}^{2}\Big(\boldsymbol{J}_{T,i}^T\,\boldsymbol{F}_{gr,i} + \boldsymbol{J}_{R,i}^T\,\boldsymbol{M}_{gr,i}\Big) + \sum_{k=1}^{12}\boldsymbol{J}_{R,k}^T\,\boldsymbol{M}_{j,k} \quad (1)$$

The vectors $\boldsymbol{q},\dot{\boldsymbol{q}},\ddot{\boldsymbol{q}} \in \mathbb{R}^{42}$ denote the generalized coordinates, velocities and accelerations, $\boldsymbol{M} \in \mathbb{R}^{42\times 42}$ is the mass matrix and accounts for the inertial properties of the system, $\boldsymbol{h} \in \mathbb{R}^{42}$ contains all gravitational and gyroscopic forces and torques. As above mentioned, muscle properties are not taken into consideration. Consequently the external forces acting on the multibody system, see figure 1, are the ground reaction forces $\boldsymbol{F}_{gr,i} \in \mathbb{R}^3$ and torques $\boldsymbol{M}_{gr,i} \in \mathbb{R}^3$ and the 12 joint torques $\boldsymbol{M}_{j,k} \in \mathbb{R}^3$, which are multiplied with the corresponding Jacobian matrices of translation $\boldsymbol{J}_T \in \mathbb{R}^{3\times 42}$ and rotation $\boldsymbol{J}_R \in \mathbb{R}^{3\times 42}$.

The equations of motion (eq. (1)) can be solved in two different ways. First, assuming given forces and torques, the movement of the body is computed using numerical integration (direct dynamics). Unfortunately, control laws to compute forces and torques of normal and impaired human walking are unknown. Therefore only the second method, called inverse dynamics, can be used. Inverse dynamics method supposes given movement and yields forces and torques.

In single support phase, when only one foot is on the ground, we deal with one ground reaction force and one ground reaction moment. Assuming given generalized coordinates and their derivatives, eq. (1) yields 42 known quantities on the left side. 3 components of $\boldsymbol{F}_{gr}$, 3 of $\boldsymbol{M}_{gr}$ and 3 of the 12 joint torques lead to 42 unknowns on the right side. We calculate the unknown quantities by matrix inversions and multiplications.

In double support phase, when both feet are on the ground, we deal with 2 unknown ground reaction forces and torques and 36 unknown joint torques, so that the number of unknowns (48) exceeds the number of available equations. Additional equations are delivered using optimization techniques, when we require that some objective functions has to be minimized. Since human walking is an optimized movement, not only the equations of motion have to be fulfilled, but also biological principles have to be taken into account. This method has already been applied to the walking cycle of a grasshopper by [14]. We assume the optimization criterion C linear in the squares of the unknown forces and torques:

$$\begin{aligned} C &= f\left(\boldsymbol{F}_{gr}^2, \boldsymbol{M}_{gr}^2, \boldsymbol{M}_j^2\right) = \frac{1}{2}\boldsymbol{F}_{gr}^T\boldsymbol{C}_1\boldsymbol{F}_{gr} + \\ &+ \frac{1}{2}\boldsymbol{M}_{gr}^T\boldsymbol{C}_2\boldsymbol{M}_{gr} + \frac{1}{2}\boldsymbol{M}_j^T\boldsymbol{C}_3\boldsymbol{M}_j \qquad (2) \end{aligned}$$

with the matrices $\boldsymbol{C}_i$ which weight the forces and torques.

The Lagrange multiplicators $\boldsymbol{\lambda}$ combine the criterion eq. (2) with the equations of motion in eq. (1) and lead to the Lagrangian function L

$$\begin{aligned} L = C &+ \boldsymbol{\lambda}^T\left\{\left[\boldsymbol{J}_{T,gr}^T \mid \boldsymbol{J}_{R,gr}^T\right]\begin{pmatrix}\boldsymbol{f}_{gr}\\ \boldsymbol{t}_{gr}\end{pmatrix} + \boldsymbol{J}_{R,j}^T\boldsymbol{t}_j\right. \\ &- \left.\boldsymbol{M}\,\ddot{\boldsymbol{q}} + \boldsymbol{h}\right\}. \qquad (3) \end{aligned}$$

According to Lagrange theory, a necessary condition for the minimisation of the criterion is that all partial derivatives of L with respect to the unknowns $\boldsymbol{f}_{gr}$, $\boldsymbol{t}_{gr}$, $\boldsymbol{t}_j$ and $\boldsymbol{\lambda}$ have to be zero, see [15]. This leads to the following system of linear equations

$$\begin{aligned} \boldsymbol{J}\boldsymbol{f} &= \boldsymbol{m} \\ \boldsymbol{f} &= \left(\boldsymbol{f}_{gr}^T, \boldsymbol{t}_{gr}^T, \boldsymbol{t}_j^T\right)^T \qquad (4) \end{aligned}$$

which can easily be solved for the unknown ground reaction forces $\boldsymbol{f}_{gr}$, torques $\boldsymbol{t}_{gr}$ and joint torques $\boldsymbol{t}_j$ by inverting $\boldsymbol{J}$.

4. Measurements

In the "Neurologische Klinik Bad Aibling", Germany, the disturbed gait patterns of hemiparetic patients are measured using the Optotrak System. This movement analysis system, depicted in figure 2, uses 58 active markers, which are applied on the skin of the patient. Two cameras measure the cartesian coordinates of the markers and a PC calculates relative angles between joints, using geometric relations. Joint velocities and accelerations are computed by differentiating twice joint angles and filtering with a 4th order Butterworth filter. The heel and toe markers are employed to determine stance and swing phase of the legs and thus single and double support phase.

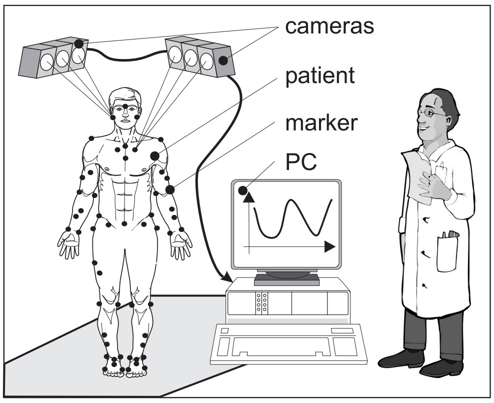

Figure 2: Measurement of human gait by a 3D movement analysis system

5. Verification

Calculated ground reaction forces and joint torques show good agreement with measurements and calculations from literature. As an example, figure 3 shows the calculated ground reaction force of a healthy subject during the single and double support phase of the right leg in comparison to data from other authors. The abscissa is time, normalized with the duration of the gait cycle and expressed in percent. A gait cycle starts with right heel contact (0%), the double stance phases are between 0% and 10% and between 50% and 60%. From 10% to 50% only the right foot and from 60% to 100% only the left foot is on the ground. The ordinate in figure 3 is the ground reaction force normalized with body weight. The calculated ground reaction force in vertical direction (bold solid line in figure 3) shows good agreement with measured data by [16] (dotted line) and calculations done by [17] (solid line).

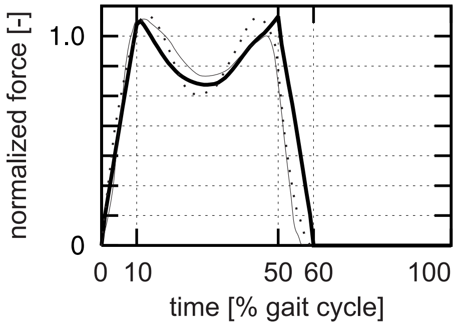

Figure 3: Vertical ground reaction forces by [16] (dotted line) and [17] (solid line) in comparison with simulation results (bold solid line)

6. Results and Conclusions

The following sections show how the method of inverse dynamics is used in the above-mentioned neurological clinic for analysis of hemiparetic gait.

6.1. Animation

Movement of the patient is automatically animated with "XAnimate" [18]. The animation offers many advantages over common recording techniques like video tapes. The doctor or physical therapist can easily, quickly and at each time view the disturbed gait pattern under different angles and zoom on single segments. Furthermore the disturbed and a normal gait pattern can be opposed on the same screen and thus differences are quickly recognised. Animation of the body's center of mass reveals excessive lateral movements of the patient. Two sample pictures of the animation of a healthy subject and a patient with severe left-sided hemiparesis are depicted in figure 4.

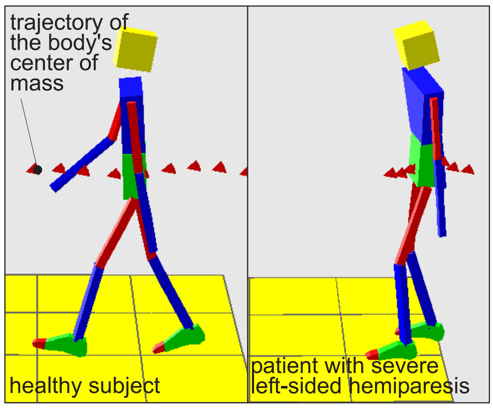

Figure 4: Animation of normal and hemiparetic gait

6.2. Evaluation of Degree of Hemiparesis and Outcome of Therapy

Calculated kinetic quantities are used to quantify the degree of hemiparesis of the patients and to control outcome of therapy.

Figure 5 shows for example peek ankle plantar flexion torque for a healthy subject A, patients with moderate (B) and severe (C and D) hemiparesis. One realises a close correlation between the degree of hemiparesis and this kinetic quantity. The more normal the gait pattern is, the higher and the more equal torques on the right and left side are.

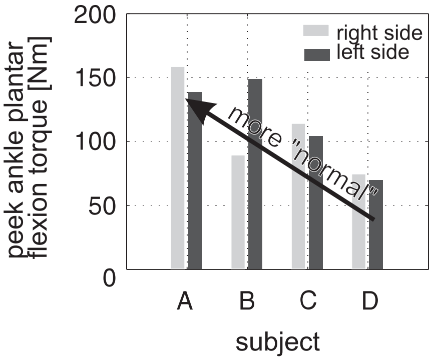

Figure 5: peak ankle plantar flexion torque for 4 subjects

Figure 6 depicts the calculated mechanical work in ankle for two steps for a hemiparetic patient at the beginning (D1), after 3 weeks (D2) and 7 weeks (D3) of physiotherapy in comparison to a healthy subject (A). One states an evident evolution of very low values of flexion work at the beginning to more normal values in course of rehabilitation. Here the outcome of therapy is evaluated by the method of inverse dynamics.

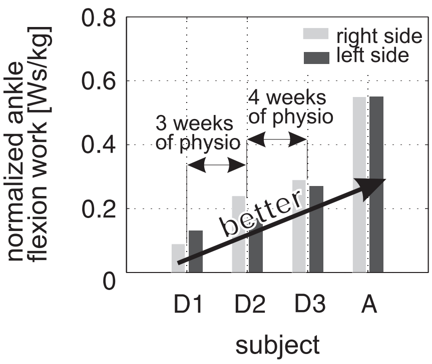

Figure 6: Normalized ankle flexion work in course of rehabilitation (physio=physiotherapy)

6.3. Detailed Analysis of One Patient's Gait Pattern

Inverse dynamics method yields time course of all kinetic quantities for one gait cycle. These data are the base for a detailed analysis of the impaired gait. Time course of left knee flexion/extension torque for a healthy subject (solid line in figure 7) and a patient with severe left-sided hemiparesis (dotted line in figure 7) are for example studied. The patient exhibits in single support phase of the left leg (between 60 and 100 % in gait cycle) a high and constant knee extension torque, whereas for the healthy subject the torque decreases very quickly to low values. Figure 8 shows the healthy subject and hemiparetic patient at 80 % in gait cycle. In single support phase the left knee of the healthy subject is completely extended.

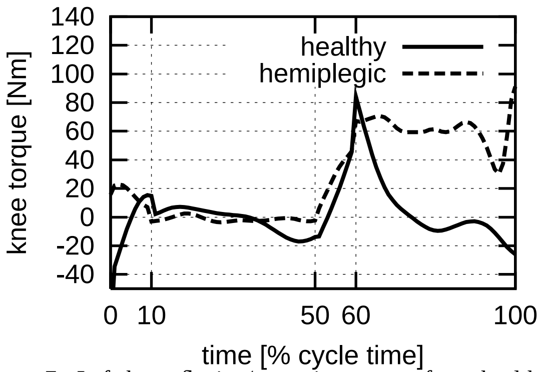

Figure 7: Left knee flexion/extension torque for a healthy subject (solid line) and a hemiplegic patient (dotted line)

Due to the intact neural control of walking, the body is well controlled falling forward in order to reduce knee torques. In contrast to this perfect neural control of movement, the hip and knee of the patient, see figure 8, are flexed, neural control of movement is impaired and the patient has to generate a very high extension torque at the knee to maintain body upright and to progress. Here kinetic quantities give insight into impaired neural control of movement.

6.4. Kinematic Characteristics of Hemiparetic Gait

Furthermore it was found that hemiparetic gait is characterised by excessive lateral movement of body's centre of mass, high lateral ground reaction forces on affected side, low peak torques in all joints of the lower extremity in comparison to healthy subjects, lower peak torques on the affected than on the unaffected side and that energy for propulsion of the body results mainly from the unaffected side.

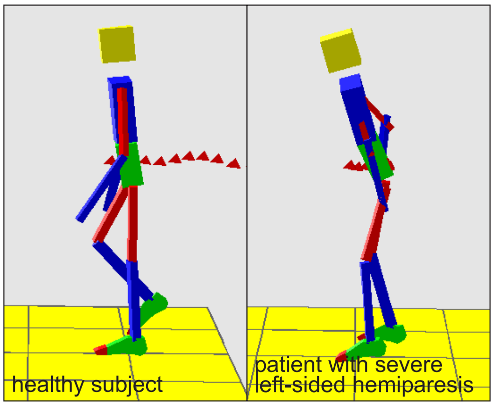

Figure 8: Animation of the two subjects of figure 7

7. Future research work

Future research work is directed towards therapy of hemiparetic patients by model-based functional electrical stimulation (FES), see figure 9.

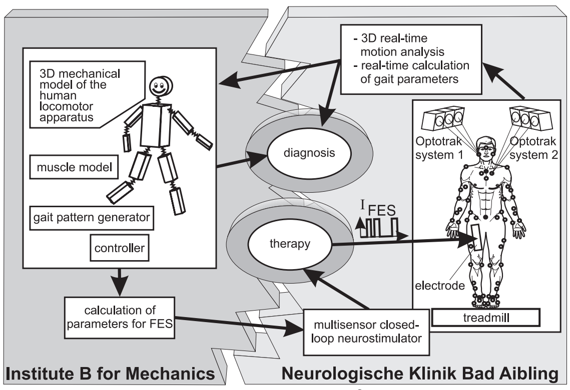

Figure 9: Diagnosis and therapy of hemiparesis: project overview

8. Summary and Conclusions

The goal of clinical gait analysis is to analyse the impaired gait pattern of the patient and to define a suitable therapy. In common gait analysis, only the kinematic data of the gait pattern are analysed, although kinetic quantities enable a further refinement of diagnosis and a better understanding of pathological gait. This paper presents a general method, derived from robotics, to calculate non-measurable kinetic quantities of human gait. In contrast to methods described in literature, expensive force platforms are superfluous. The calculation of kinetic quantities is based on a mechanical model of the human locomotor apparatus. The Newton-Euler method yields the equations of motion. Using inverse dynamics and optimisation, ground reaction forces and joint torques are computed. Movement as input into the inverse mechanical model is measured with a contactless movement analysis system. Calculated kinetic quantities show good agreement with data from literature. The method is applied in the "Neurologische Klinik Bad Aibling", Germany for analysis of hemiparetic gait. Animation of the impaired gait pattern yields a better visualisation than video recordings. The degree of hemiparesis and outcome of therapy is evaluated and patient's gait is detailed analysed by kinetic quantities. More generally, a characterisation of hemiparetic gait by kinetic quantities was found. Future research work deals with therapy of hemiparesis by functional electrical stimulation.

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