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An application of CaTraSys, a cable-based parallel measuring system for an experimental characterization of human walking

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SUMMARY

In this paper, an application is presented of a cable-based parallel manipulator as measuring system for an experimental identification of human walking characteristics. Experimental results have been obtained by means of a new version of CaTraSys (Cassino Tracking System), which is a measuring system that has been designed and built at Laboratory of Robotics and Mechatronics (LARM) in Cassino, Italy. The new version of the CaTraSys system has been used to determine the trajectory of the human limb extremity during walking operation and furthermore the system is able to measure forces that are exerted by a limb. Experimental determination of articulation mobility is also presented with numerical and experimental results.

KEYWORDS: Experimental robotics; Parallel manipulators; Cable-based architectures; Measuring systems; Medical applications.

1. Introduction

In general, the human gait has been studied through visual observations and in recent years, by combining advanced measurement technology and biomechanical modeling. For rehabilitation of an articulation, mobility and force/torque measure can be used not only in a preliminary stage but also during the rehabilitation therapy to get a measure of improving capability of a human subject.[1] Furthermore, the technology and design of joint implant systems and human rehabilitation have evolved considerably with a variety of new systems that are available in the market. Clinical and radiological outcome criteria have been well established. However, for example, a functional comparison of different available knee arthroplasty systems is difficult because of numerous patient-related variables and varying expectations of patients.[2]

Therefore, it is of great interest to develop and use measuring systems, which can be used for diagnosis applications. Common tools are:[1] optical devices, such as photography, stroboscopy, cinematography, videography, and magnetic resonance. Goniometry and electro-goniometry give joints kinematic data, both for static and dynamic actions. Electromyography gives muscle action potentials. Dynamography and accelerometer sensing give direct or indirect force measures. Computer-aided simulation gives an overall prediction of kinematic and kinetic performances of patient movement. A six-axis force-torque sensor can be used to get information about wrenches and a 6 axis position-orientation sensor can be used to measure a displacement of a patient. For example, a measuring system that is available in the market with these features is the so-called Qualisys Motion Capture System (QMCS),[3] which offers a product for end-users to analyze the gait. It can be used even for clinical research studies.

Economical obligations and needs for an easy implementation in medical environment stimulate a demand of alternative measuring systems or proper adaptation and redesign of existing devices.[1]

In this paper, we have addressed the attention to cable-based tracking systems, which appear to be interesting since they show a good compromise between accuracy, resolution, cost, measurement range, portability, and calibration procedure. Cable- based systems are portable; therefore, they can be brought on site (i.e., hospital environment and even patient's home), they are of low cost and easy to use, and once the calibration procedure is performed it is not necessary to repeat it before the use. Moreover, they can have a very large workspace for measuring moving objects. In particular, in this paper we have adapted an existing system, which was successfully used for evaluating robot characteristics[4–6] to the analysis of human walking characteristics.

A cable-based tracking system consists of a fixed base and a moving platform that is connected by at least six cables whose tension is maintained by pulleys and spiral springs in the base. They can be modeled as 6-degrees of freedom (DOF) parallel manipulators because cables can be considered as extensible legs connecting the platform and base by means of spherical and universal joints, respectively. One drawback of this kind of measuring devices is that they are forced to operate in a reduced workspace, in order to avoid cables interference.

Furthermore, cables tensions must be bounded since they must have a minimum value to avoid cables to become slack, and a maximum value to minimize elastic and inelastic effects.

Cable-based measuring systems can be used for a determination of the articulation motion and walking capabilities of a patient. Therefore, they can be used in a preliminary stage of rehabilitation therapy to get information about the motion capabilities of a patient.

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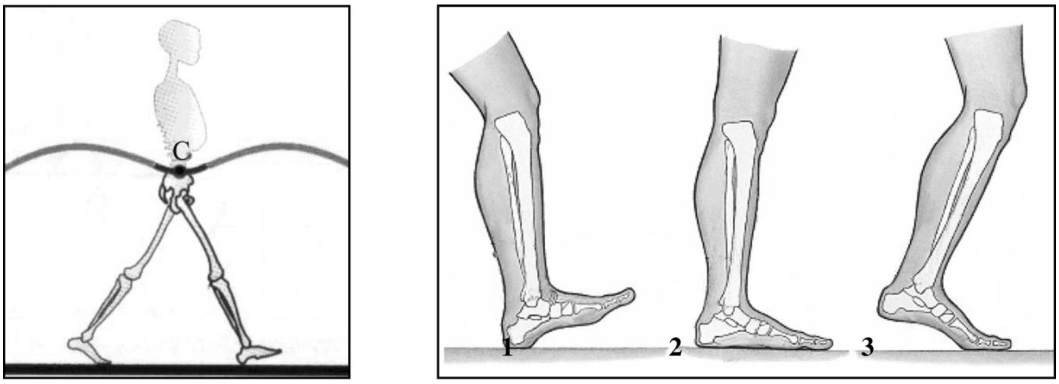

(a) (b)

Fig. 1. Support phases of the human walking: (a) double support; (b) detail on the phase support: (1) refers to heel contact, (2) is sole contact, and (3) refers to tip contact.

Cassino Tracking System (CaTraSys), which has been designed and built at Laboratory of Robotics and Mechatronics (LARM) in Cassino, is a pose measuring device,[4] and it has been considered in this paper for the above-mentioned application. This measuring system has been firstly used for the experimental evaluation of robot kinematic performances.[4–6] In particular, a numerical procedure has been used in ref. [4] to determine H-D parameters of the kinematic chain of industrial robots, as reported in. ref. [7]. Then, the procedure has been specifically adapted for experimental determination of kinematic parameters of human limbs.[8] A human leg has been schematically modeled by using two revolute joints for the hip mobility and one revolute joint for the knee mobility in order to measure main kinematic characteristics.

In this paper, a new version of the CaTraSys measuring system has been proposed for diagnosis purposes. It allows pose and wrenches measurements. Preliminary results of the new measuring system have been presented in refs. [8, 9]. A preliminary version of this paper has been presented at MUSME08, IFToMM-FeIbIM International Symposium on Multibody Systems and Mechatronics in April 2008.

In this paper, a Kinetostatic model is proposed to evaluate wrenches that are applied to the end-effector for CaTraSys as function of cables tension. The Kinetostatic model has been proposed for the 6-DOF measuring system and then for a simplified system for position and force analysis with 3 DOF. An experimental validation of the proposed model is performed by means the new version of CaTraSys, in which load sensors have been considered. In particular, the system has been used for the experimental kinetostatic analysis of human walking. Two types of experiments have been performed: the first one is the force and motion range determination for each leg articulation; the second one refers to kinetostatic performances of human walking. Force sensing can be important for diagnosis purposes, when it is advisable to monitor improvements and recovering both in terms of mobility and force capability. Therefore, a comparison is also presented between numerical and experimental results. In this paper the system has been presented for diagnosis purposes, but it can be used for further application as rehabilitation device since it can exert forces and torques on the leg end point as guiding with active, passive, or force-motion evaluation modes in rehabilitation exercises. The force measuring can be considered interesting since, in the proposed application, the measured force can indicate the state of leg performance, when compared with suitable reference values of normal walking.

2. Human Walking and Rehabilitation Issues

Nature has solved efficiently the problem of biped locomotion during a long evolution of human beings. Human walking is very smooth and efficient because humans can make use of gravity effectively.[10] The energy that is needed to walk is low, and human walking motion is energy optimized.[1] Basic features that have been deduced from walking operation in steady state are related to features of approximately symmetric and periodic walking. A walking cycle can be divided into strides, which are themselves divided into steps. Each cycle is composed of single and double supports. Figure 1 shows main phases of two legs walking through double support gait, as shown in Fig. 1(a) while Fig. 1(b) shows the ground contact phases of each foot.

The walking operation can vary from one individual to another depending on gender, age, weight, and height. A same individual has various walking patterns depending on learning and tiredness, which are a source of modification of the gait. The objective of a movement also influences the gait. In order to efficiently cover a long or a short distance, the walking parameters such as step length and speed can be different.[10]

Equilibrium is a key point of all standing attitudes, for posture and/or locomotion. Postural static equilibrium such as standing at a workstation is guaranteed when the center of gravity projection on the ground is maintained within the support base. The support base is delimited by the points of the system in contact with the ground, which corresponds to the surface of the convex hull linking the contact points together.[11]

Normal walking corresponds to dynamic walking with a fall forward onto the foot receiving the body's weight; the center of gravity projection does not remain inside the support base during the whole movement. The only way to avoid

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falling is to place the swing foot on the ground, in front of the center of gravity projection.

Many biomechanics studies on human walking are available in the literature. Most of them characterize the different phases that compose the locomotion cycles. An accurate description of these phases is provided by Nilsson in ref. [2].

The study and knowledge of biomechanics of human movement can give useful information to determine the normalcy of the walking operation and function of a subject. In fact, that information can be used both for diagnosis procedures and for further rehabilitation therapies.

Rehabilitation therapies deal with the study and reactivation of movement patterns of injured and/or disabled persons, through strategies that are based on the integration of activity by doctors, physiotherapists, and medical personnel. Biomechanics can help the rehabilitation therapies in analyzing movement patterns, predicting forces acting on the joints and muscles, designing assistive devices, optimizing performance, and reassessing performance after rehabilitative training.

Therefore, rehabilitation involves several activities, namely, diagnosis, exercise, rehabilitative actions, and analysis of normal daily living movements. The goals of a rehabilitation planning can be summarized as:[1]

(1) Definition of the biomechanics characteristics of a patient, in terms of diagnosis of articulation and muscular capacity, together with motion abilities and disabilities,
(2) Evaluation of movement patterns and posture of patients,
(3) Comparison of results with normal patterns and postures,
(4) Definition of a training program and exercises, which should be developed by the help of therapists or under their guide or supervision, and
(5) Definition and design of assisting devices, which can be used together with medical personnel or not.

Exercises in rehabilitation therapies are usually carried out by slow movements of the human limb in order to reduce inertia effects and to allow the muscles to contract gradually throughout the range of motion. Furthermore, the movements of the body parts should be carried out always maintaining a certain alignment of the parts to avoid extra forces and moments in muscles, tendons, and ligaments. The balance of antagonistic muscles groups should be maintained. Finally, the definition of a set of exercises will prevent the patient to be bored and will provide versatility for home environments or self-practice. As can be easily understood, several aspects of rehabilitation activity can be devoted to assisting devices, even for self-training and home installations. Recently, attention has been addressed to the design and experimentation of new assisting devices with cable-based architectures. They may have features for transportability and easy installation, but in general they still present a complex way of patient adaptation, so that one critical issue can be recognized in human interfaces that are specifically devoted to the adaptability to each patient in term of physical and psychological conform during the therapeutic exercise.[12] Motion capture system or measuring system should be used before and during the rehabilitation therapy to monitor/verify/adjust the motion capabilities of a patient.

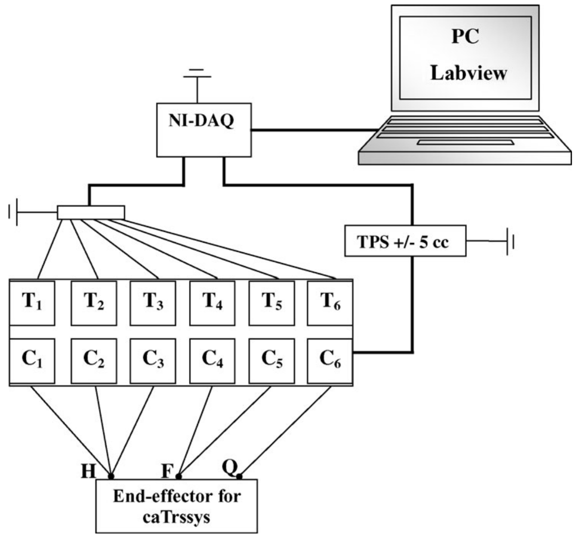


Fig. 2. A scheme of the new force-sensored cable-based measuring system CaTraSys: T_i is a cable transducer; C_i is a force sensor.

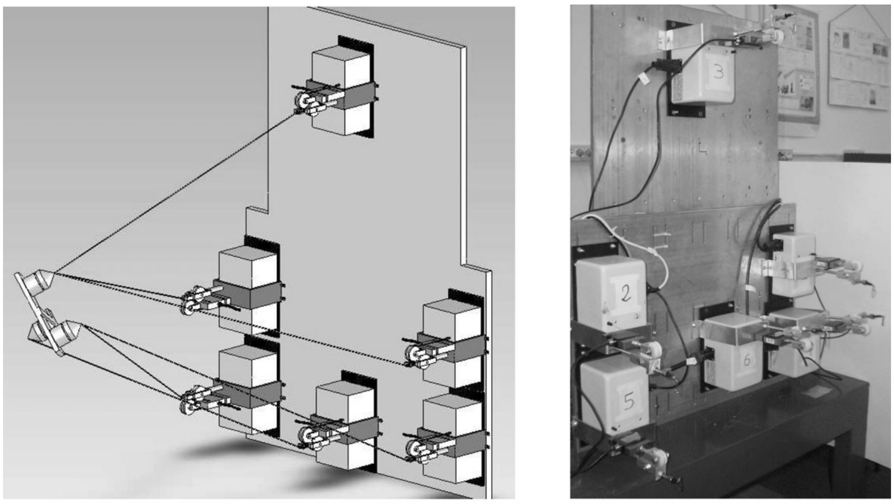
3. The New CaTraSys as Pose/Wrench Measuring Device

CaTraSys is a cable-based measuring system. It has been conceived at LARM since 1994[4] in order to identify the pose of a rigid body during a large motion through on-line computation of the Kinematics of the designed 3-2-1 cable-based parallel architecture. In this paper we present a new version of the CaTraSys, which is able of measuring and monitoring pose and wrenches, through a design that is shown in Figs. 2 and 3. It determines the pose (position and orientation) of a moving object by using trilateration technique. Details of CaTraSys are reported in refs. [4–9]. It is composed of a mechanical part, an electronics/informatics interface unit, and a software package. The mechanical part consists of a fixed base, which has been named as Trilateral Sensing Platform, and a moving platform, which has been named as end-effector for CaTraSys.

The two platforms are connected by six cables, whose tension is maintained by pulleys and spiral springs that are fixed in the base. The new version of the CaTraSys is able to measure end-effector poses and wrenches. In order to obtain this result, position transducers T_i and force sensors C_i have been used, as shown in Figs. 2 and 3.

The end-effector for CaTraSys is the moving platform operating as a coupling device since it connects the cables of the six transducers to the extremity of a moving system as it is shown in Fig. 3. It allows the cables to track the system while it moves. Signals from cable transducers are fed though an amplified connector to the electronic interface unit, which consists of a laptop for data analysis.

In this paper, a modified version for CaTraSys measuring system is considered as an enhancement of what has been preliminarily presented in ref. [9], in which force sensors can be suitably used to obtain both pose and wrenches information. In the mechanical design in Fig. 4 each force



(a) (b)

Fig. 3. A design for the new CaTraSys: (a) a CAD scheme, (b) a laboratory lay-out.

sensor has been installed in the fixed platform with two pulleys. The proposed solution gives the advantage of reducing inertial effects (which are due to the cables only) and there is no need to have miniaturized force sensors but commercial ones can be used. The pulleys have been sized to have a compact system with a large orientation capability for each cable and to avoid the risk of cable folding/damaging.

Figure 4 shows a scheme and suitable installation of the force sensors in the CaTraSys prototype. The resulting system for tension monitoring is compact and can be easily adapted to other cable tracking systems and cable-based parallel manipulators.

4. A Kinetostatic Analysis

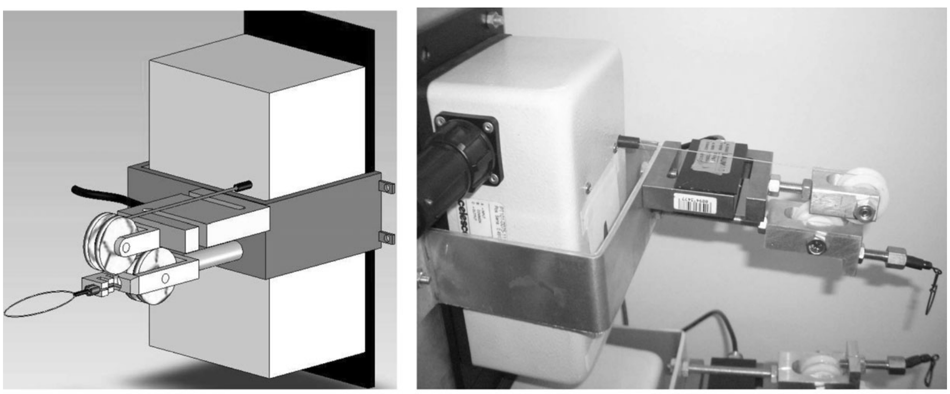
Cable-based parallel architectures deal with the problem of exerting or sensing forces and torques. Indeed, kinematic analysis is important, but also static equilibrium is a fundamental issue. In passive cable-based manipulators, wrenches must be bounded in order to limit the influence of the measuring device on the system under measure.

The static equilibrium of a cable-based parallel manipulator can be formulated referring to the model in Fig. 5 as

$$\sum_{i=1}^{n} \mathrm{F}_i = -\sum_{i=1}^{n} F_i \hat{\mathrm{l}}_i = \mathrm{P}; \quad \sum_{i=1}^{n} \hat{\mathrm{l}}_i \times R\mathrm{b}_i = \mathrm{T}. \tag{1}$$

In Eq. (1), $\mathbf{F}_i$ is the cable tension that is applied to the ith cable and it is in the negative direction of the cable length unit $\mathbf{l}_i$ because $\mathbf{F}_i$ must be in tension. R is the rotation matrix relating the orientation of the moving frame HX′Y′Z′ to the fixed frame OXYZ, as shown in Fig. 5(a). The unit vectors of the moving frame are expressed by **i**, **j**, and **k**. Moreover, $\mathbf{b}_i$ (for $i = 1, \ldots, 6$) are the position vectors from H to the corresponding ith cable attachment points, as expressed in the moving frame; and **P** and **T** are the resultant vector force and torque (when considered together they give a wrench **W**) that are exerted on or by the environment, in accordance with the active or passive nature of the cable-based system. Substituting the above-mentioned terms in Eq. (1) yields to

$$J^T \mathrm{F} = \mathrm{W}, \tag{2}$$



(a) (b)

Fig. 4. A zoomed view for a tension monitoring system that is installed in the new CaTraSys in Fig. 5: (a) a three-dimensional sketch, (b) laboratory installation.

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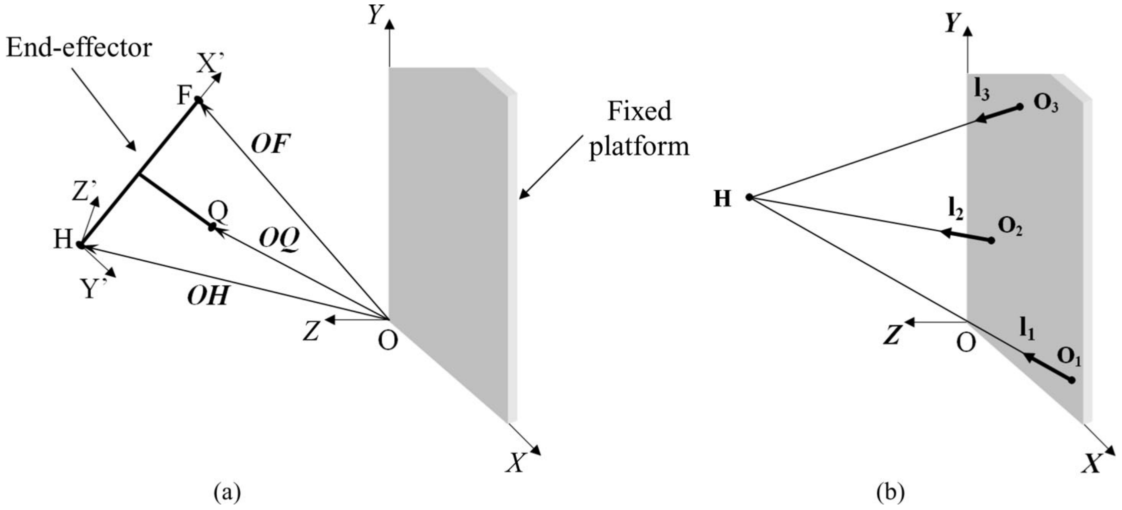

Fig. 5. A scheme for the kinetostatic analysis of CaTraSys: (a) a 6 cables' model, (b) a 3 cables' model.

in which $F = [\, F_1 \cdots F_n \,]^T$ represents the vector of n scalar cable forces; **W** is the resultant external end-effector wrench that is expressed in the fixed frame; J^T is the transpose of the Jacobian matrix. In this case the manipulator Jacobian matrix is [6×6] but in general, for cable-driven parallel manipulator it is [m×n], where m is number of controllable end-effector DOFs and n is the number of cables. The kinematics of CaTraSys system has been solved analytically as due to its trilaterable nature by using trilateration as reported in refs. [4, 5] and Cayley-Menger determinants, as explained in ref. [13]. Hence by applying three consecutive trilateration operations to determine the locations of points H, F, and Q in Fig. 5(a), the orientation of the moving frame with respect to the fixed one can be expressed as

$$\mathrm{i} = \frac{\mathrm{FH}}{\|\mathrm{FH}\|}; \quad \mathrm{j} = \mathrm{k} \times \mathrm{i}; \quad \mathrm{k} = \frac{\mathrm{FH} \times \mathrm{QH}}{\|\mathrm{FH} \times \mathrm{QH}\|}. \tag{3}$$

Unit vectors **i**, **j**, and **k** can be used to determine the R matrix in Eq. (1). Figure 5(b) refers to a simplified model of the measuring system, which can be used for a position estimation device with three cables only.

In passive cable-based manipulators, it is of great interest to be able to monitor the end-effector wrenches as a consequence of the cables tensions, which can be in general described by

$$\mathrm{F_i(i)} = \mathrm{s_o(i)} + \mathrm{s(i)}\ 1\mathrm{(i)}, \tag{4}$$

in which s(i) (for $i = 1, \ldots, 6$) is the relationship between the ith cable tension and cable length l(i); and s_0(i) depends on the spring preloading for each cable transducer.

In particular, different type functions have been tested to correctly use the model in Eq. (4), as reported in ref. [8]. First, it has verified that F_i(i) is not a constant value but it depends on the cable's length l(i).

In particular, the forces exerted by the cables are required to keep them in tension. Therefore, their range is limited to practical bounds. The analysis of force variation has been carried out as function of speed and testing person.[9] The proposed models have been tested experimentally by using the new version of CaTraSys. The results of the experimental analyses that are reported in ref. [8], show that the best model is a linear relationship with s_0(i) and s(i) kept as constant values.

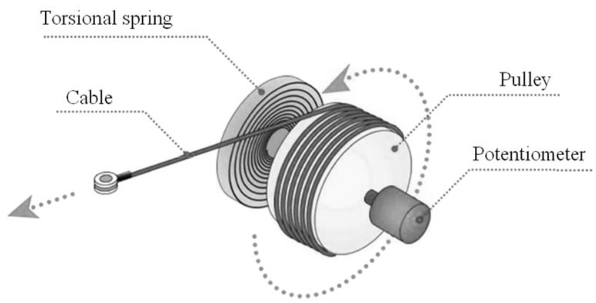

Fig. 6. A scheme for a cable transducer.[14]

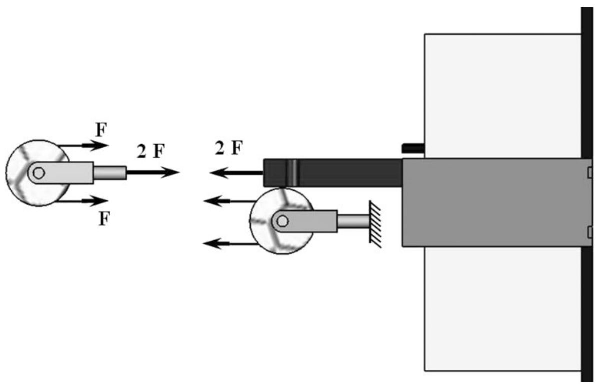

Fig. 7. A scheme for the force sensor installation with acting force.

Figure 6 shows a schematic representation for each cable transducer. It is composed by a pulley, a potentiometer, and a torsion spring, which are mounted on a common shaft. The spring must be sized in order to guarantee a suitable positive tension in the cable during its forward movement and return. Figure 7 shows a scheme for the free-body diagram of the force sensor according to the mechanical design with pulleys in Fig. 4.

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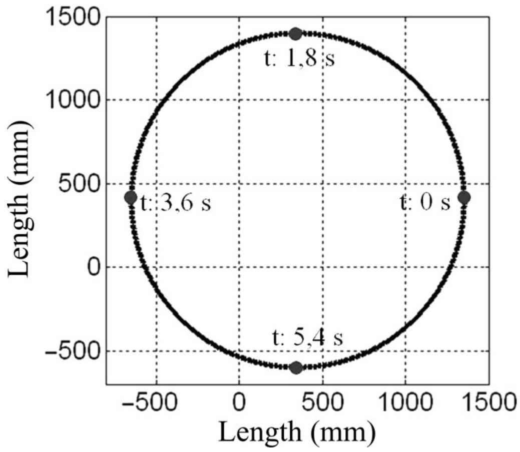

Fig. 8. A simulated circular trajectory in X–Y plane with characteristic points as function of motion time.

A simplified model for a position and force estimation device can be derived for CaTraSys with only three cables, as shown in the scheme in Fig. 5(b). In particular, once the position of H is obtained with the formulation as described in refs. [4, 5], the directions of the cables tensions coincide with the unit vectors of the cables' lengths. The tension in each cable is then evaluated by Eq. (4). The resulting force **F** is then the sum of the three cables forces F_i. In this paper we have considered the force **F** that a human operator exerts on CaTraSys.

5. Numerical and Experimental Results

A numerical simulation has been carried out as based on the formulation in Eq. (4) together with the Kinematics in ref. [5] with the aim to validate the proposed model and to size the design and operation of CaTraSys application. In particular, a circular trajectory has been planned in the X–Y plane to test and validate the kinetostatic model with three cables. Figures 8–10 show plots for the results of numerical simulation. In particular, in Fig. 8 a circular trajectory is shown in which four points are considered. Figure 9 shows the correspondent cables' lengths and forces acting in the cables. Figure 10 shows the X, Y and Z components and its magnitude of the resulting force, which is applied to the end-effector. The proposed numerical simulation can be used for further developments of the new CaTraSys for proposed monitoring of human walking. In fact, once that the kinetostatic model is validated, it can be used to identify different regions of the workspace as function of the force that can be applied by/on the end-effector. It can be also used to predict the force that is sensed by a user in a diagnosis test. Almost all the reported experiments have been carried out by considering the human motion in a saggital plane, which is parallel to the CaTraSys fixed base, so that it is possible to obtain the contribution from different planes and even as from the patient. However, the measured force by CaTraSys is always the balance with the action by the monitored patient. Calibration of the measuring system was performed before the experimental test.

Several tests have been carried out even with repeated tests for statistical validation. Measurement errors, noise

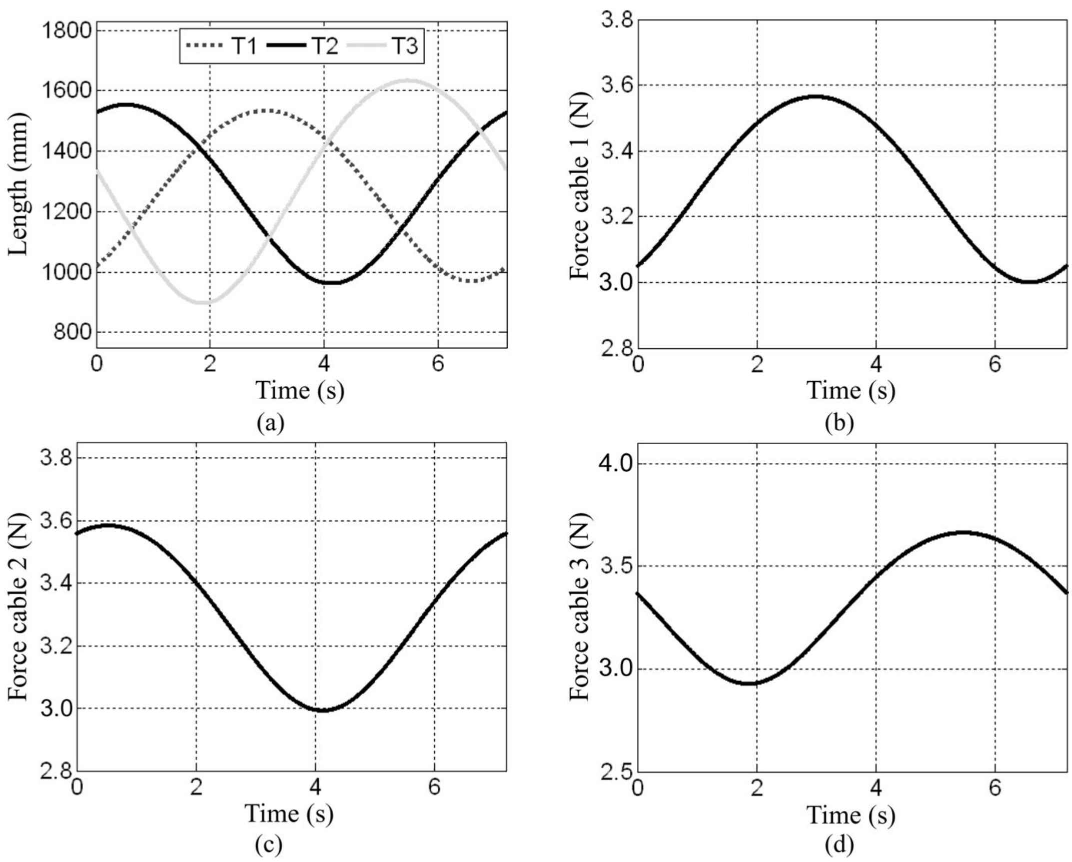

Fig. 9. Numerical results from a kinetostatic analysis of a circular trajectory: (a) lengths of cables; (b) tension in cable 1; (c) tension in cable 2; (d) tension cable 3.

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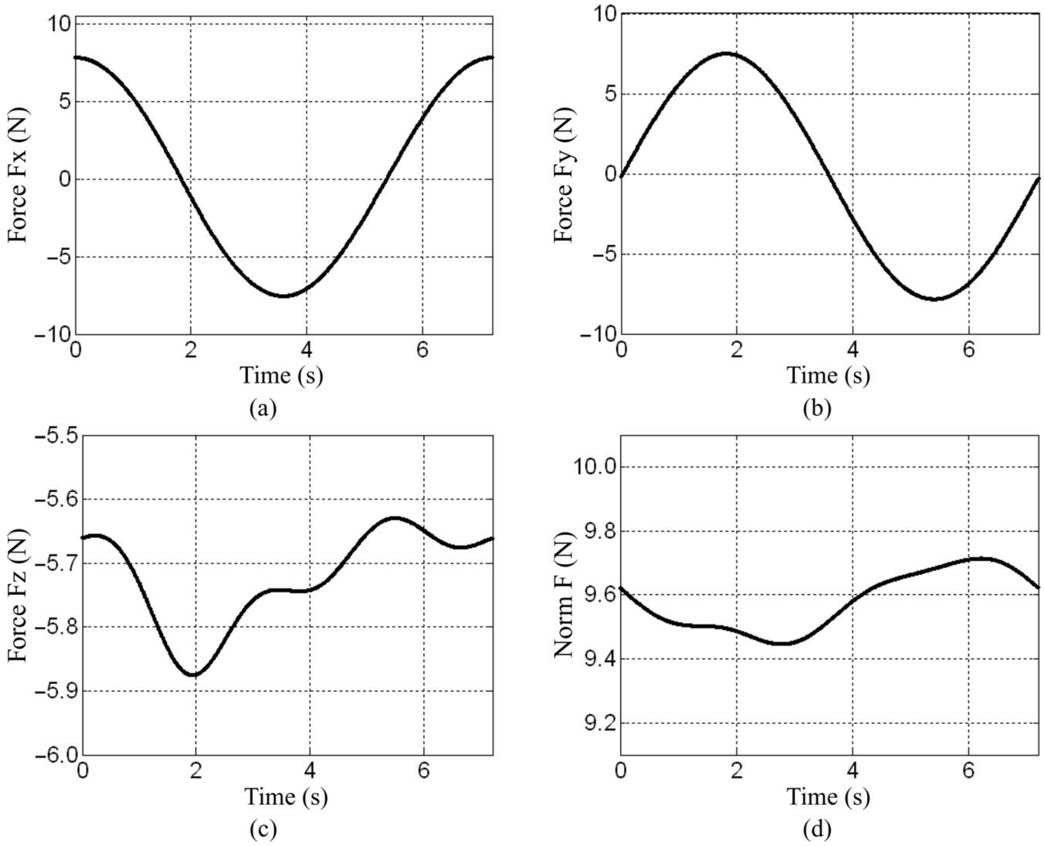

Fig. 10. Numerical results from the kinetostatic analysis of a circular trajectory in Fig. 8 in terms of force acting at the connection point: (a) X component; (b) Y component; (c) Z component; (d) magnitude.

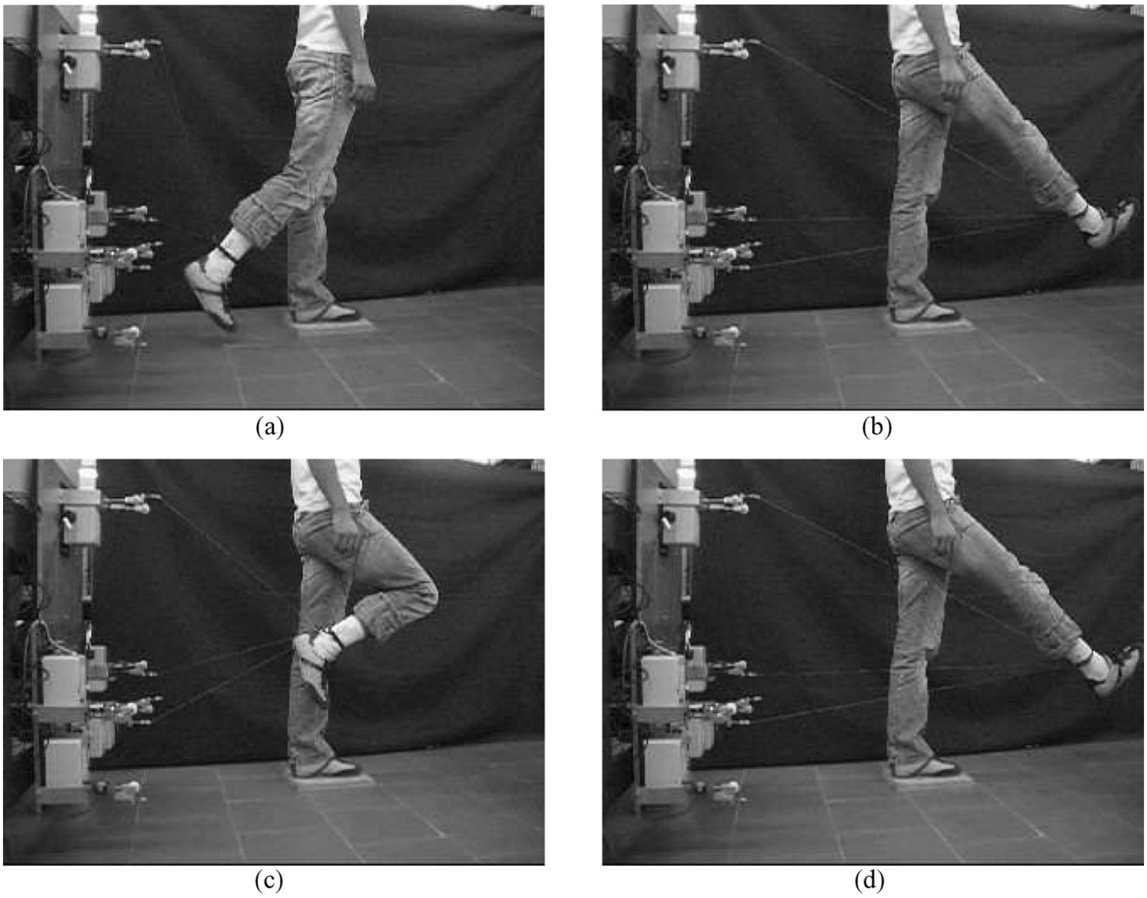

Fig. 11. A sequence of human leg movements: (a) start position for leg rotation; (b) end position for leg rotation; (c) start position for knee rotation; (d) end position for knee rotation.

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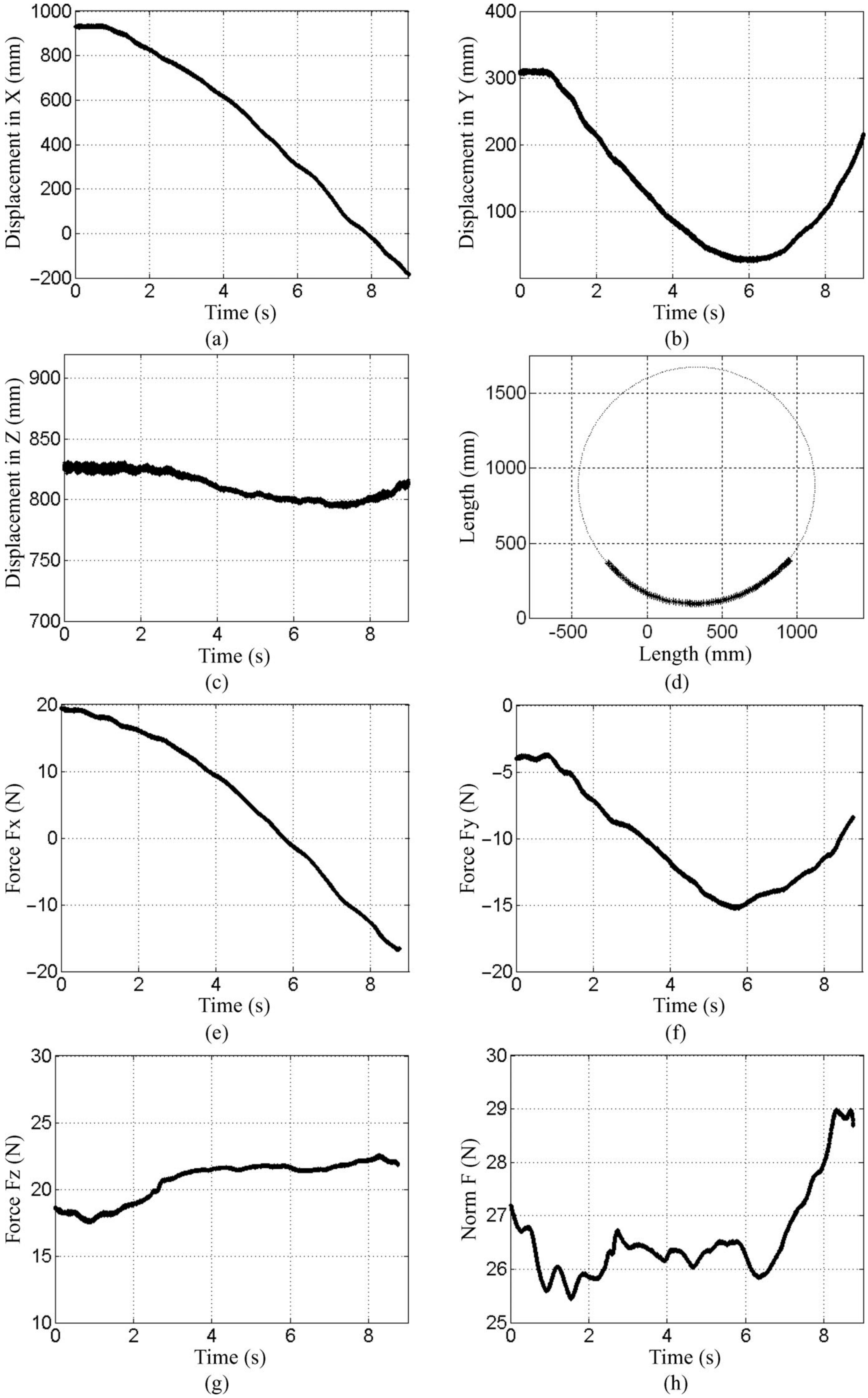

Fig. 12. Experimental results of the kinetostatic analysis of human leg movement in Fig. 11 (a) and (b) in terms of: (a) X displacement component; (b) Y displacement component; (c) Z displacement component; (d) point trajectory; (e) X force component; (f) Y force component; (g) Z force component; (h) magnitude of force.

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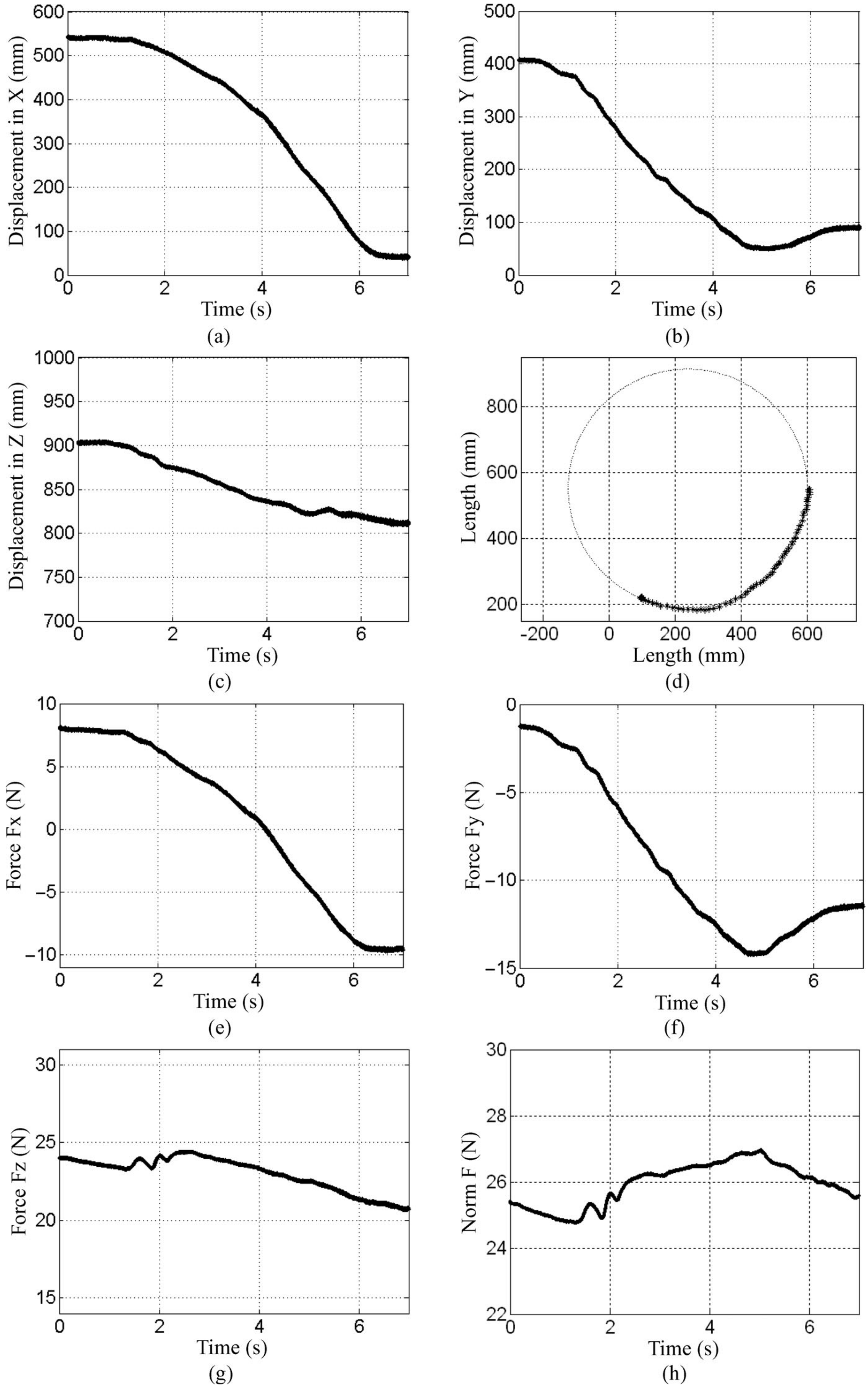

Fig. 13. Experimental results from the kinetostatic analysis of a human knee movement in terms of: (a) X displacement component; (b) Y displacement component; (c) Z displacement component; (d) trajectory of the leg end point; (e) X force component; (f) Y force component; (g) Z force component; (h) magnitude of force.

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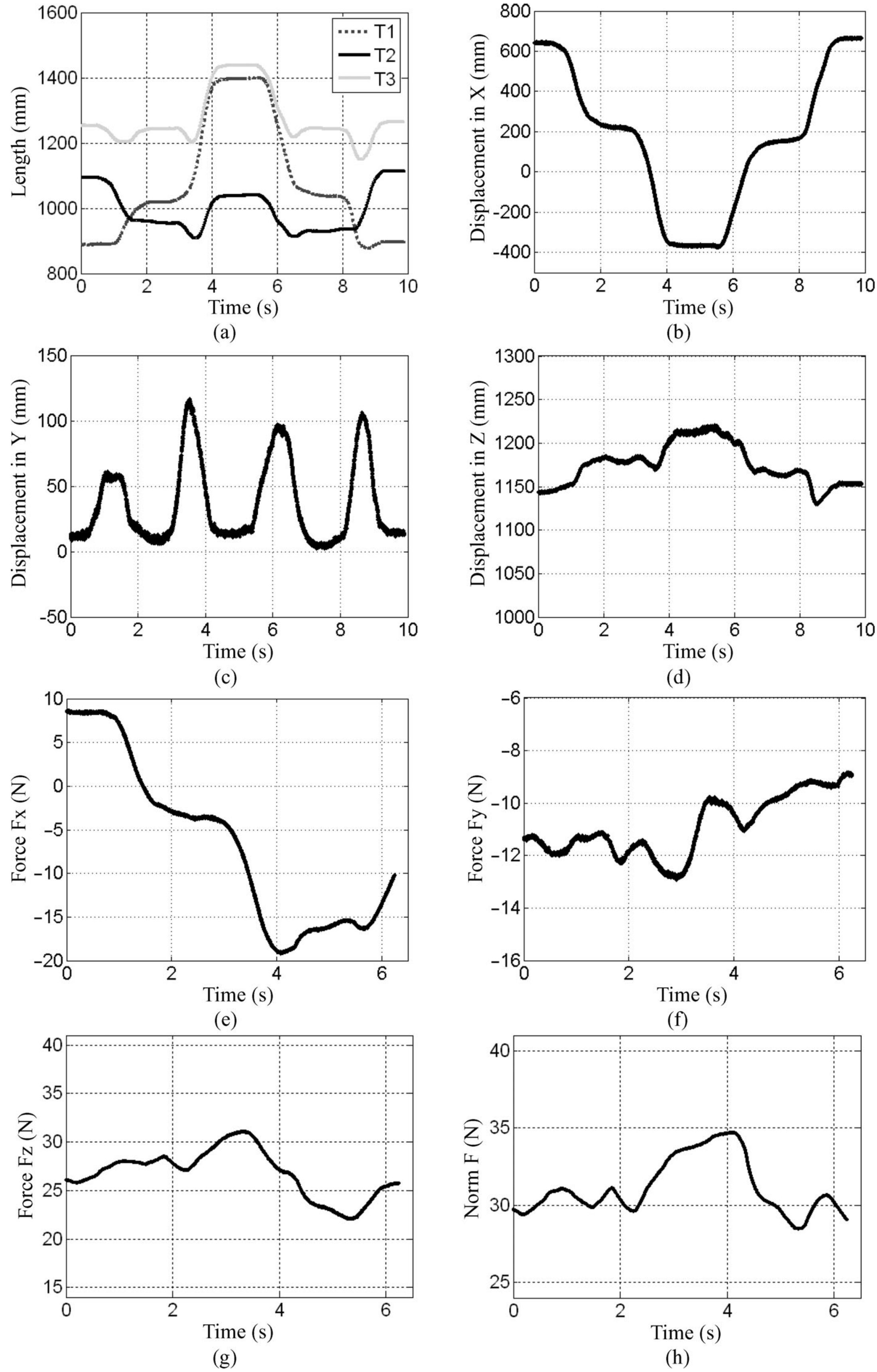

Fig. 14. Experimental results from the kinetostatic analysis of a human walk movement in the X–Y plane in Fig. 11(c) and (d) in terms of: (a) cable lenghts; (b) X displacement component; (c) Y displacement component; (d) Z displacement component; (e) X force component; (f) Y force component; (g) Z force component; (h) magnitude of force.

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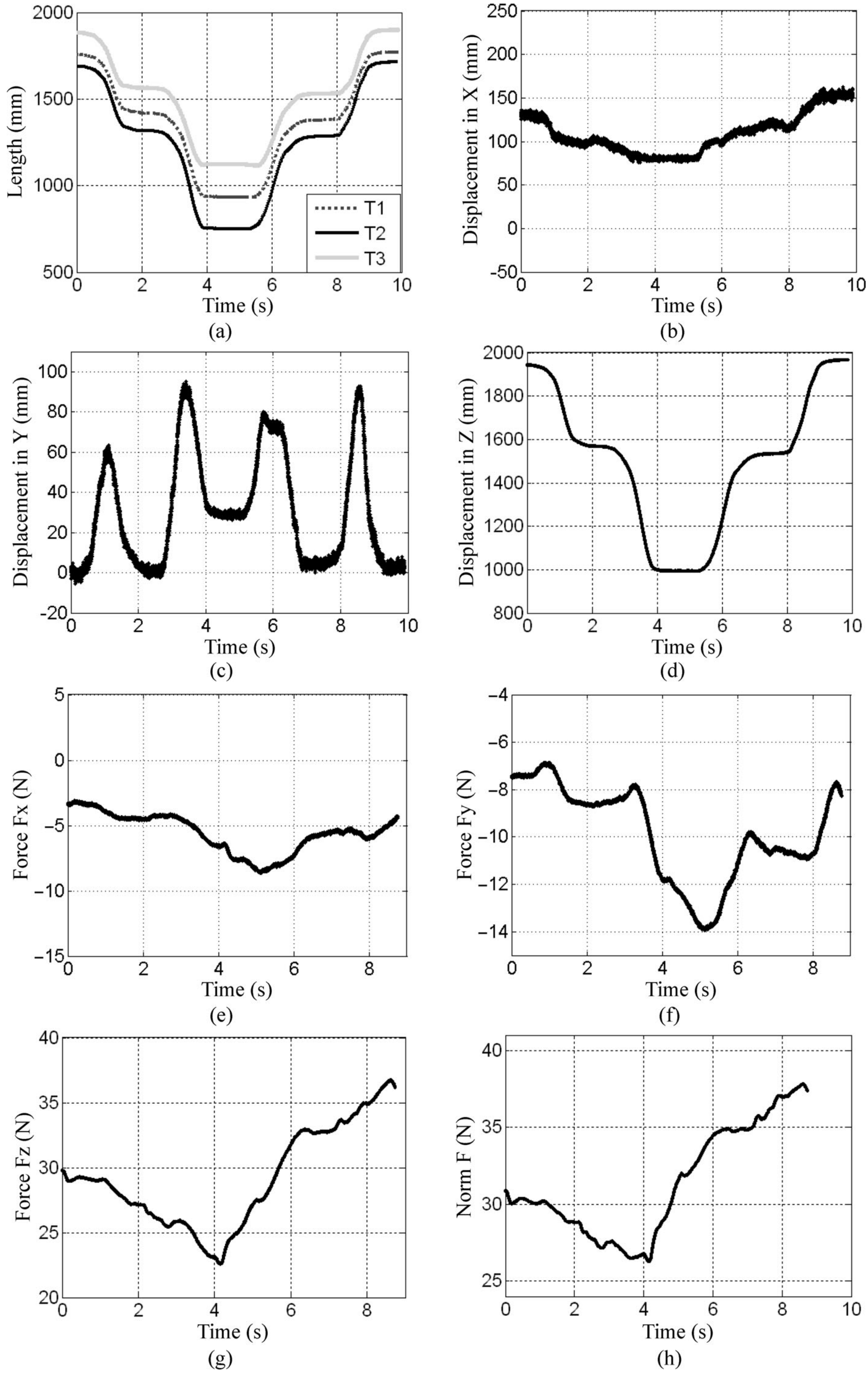

Fig. 15. Experimental results from the kinetostatic analysis of a human walk movement in the Z–Y plane: (a) cable lenghts; (b) X displacement component; (c) Y displacement component; (d) Z displacement component; (e) X force component; (f) Y force component; (g) Z force component; (h) magnitude of force.

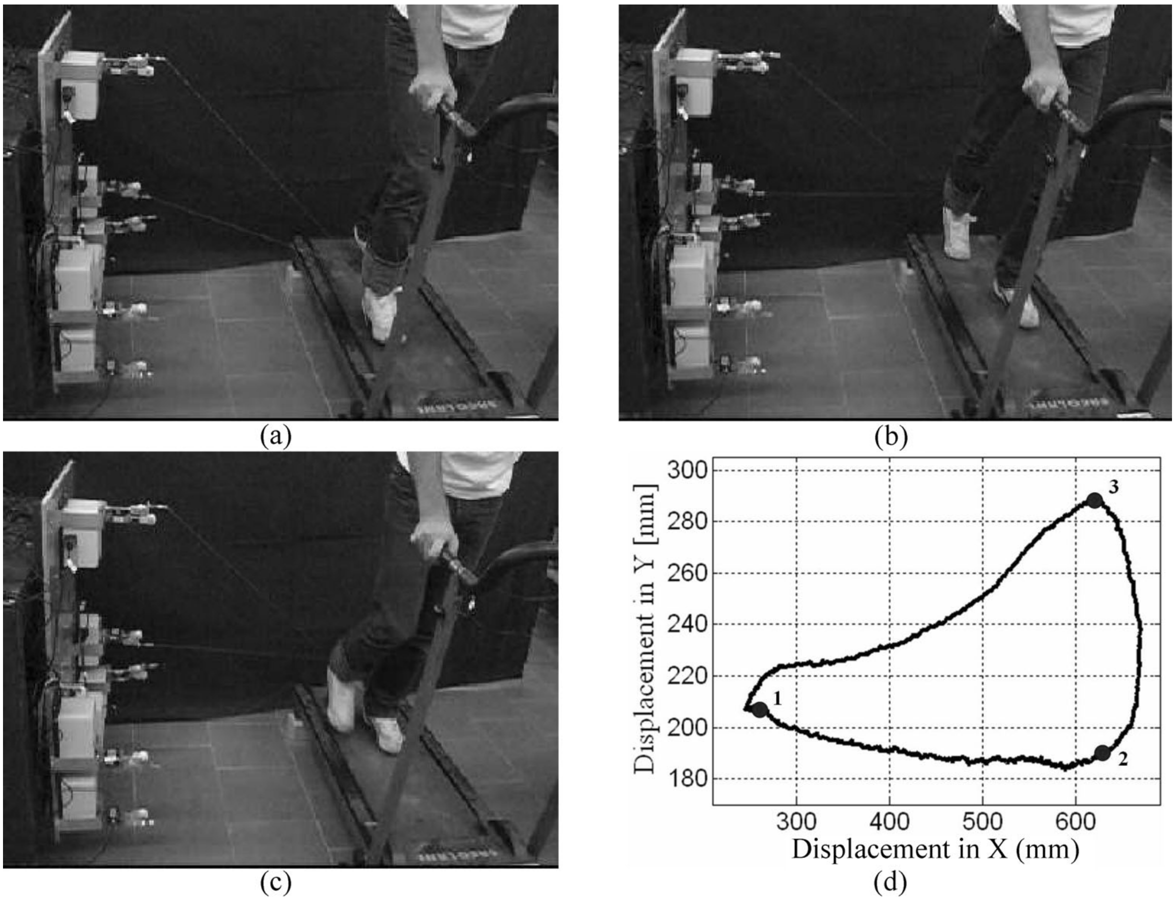

Fig. 16. A sequence for a human gait in a continous walking: (a) Position 1; (b) Position 2; (c) Position 3; (d) a trajectory of the leg end point on the X–Y plane.

and accuracy of the measuring system CATRASYS have been investigated and presented in previous papers.[5,13] In particular, CATRASYS shows an average value of accuracy of 1 mm for point displacements in a range between 1 and 2 m, when using CELESCO cable transducers.

For each experiment we have considered a scan rate of 1000 acq/sec. In particular, two types of experimental tests have been considered. The first one is related to the determination of the kinetostatic characteristics of an articulation in terms of motion/force ability. Experimental tests on leg movements have been reported in Figs. 11–13. Experimental tests have been carried out on human limbs of seven different male and female subjects by using the proposed formulation and the new version of the system. The experimental tests have been settled in the lab with constant environmental conditions for subject comfort in terms of temperature and with same test schedule.

The second type of analysis is related to the experimental determination of human walking in terms of forward and backward motion, as shown in Figs. 14 and 15, and with the aid of a tapis-roulant, as shown in Figs. 16–18.

Figure 11 shows a sequence of an experimental test on a male subject during a leg movement in saggital plane. Illustrative experimental results on three male subjects are summarized in Table I. A formulation to determine the kinematic parameters and joint ranges of human limbs has been proposed in refs. [8, 9]. In particular, a human leg has been schematically modeled by using two revolute joints for the hip mobility and one revolute joint for the knee mobility. A procedure that is described in refs. [8, 9]. has been used to determine the interpolation of a circumference resulting by a single joint movement for each link (femur and tibia) of the limb model. In Table I "r" represents the radius of this interpolating circumference; E_m and E_{max} are the average and maximum deviations among experimental data and coordinates of interpolating circumference. Angle α is the range of each joint movement. Figure 12 shows experimental results for a kinetostatic analysis of a human leg

Table I. Kinematic parameters of human legs experimentally determined with tests in Fig. 11.

Human Subject	1	2	3
Link 1: femur			
r (mm)	787.1	720.9	759.5
E_{max} (mm)	8.9	29.9	11.2
E_m (mm)	3.0	13.5	4.0
α (deg)	98.7	133.3	64.3
Link 2: tibia			
r (mm)	362.1	313.4	289.9
E_{max} (mm)	11.9	14.1	8.8
E_m (mm)	3.6	5.1	2.5
α (deg)	112.4	91.4	107.3

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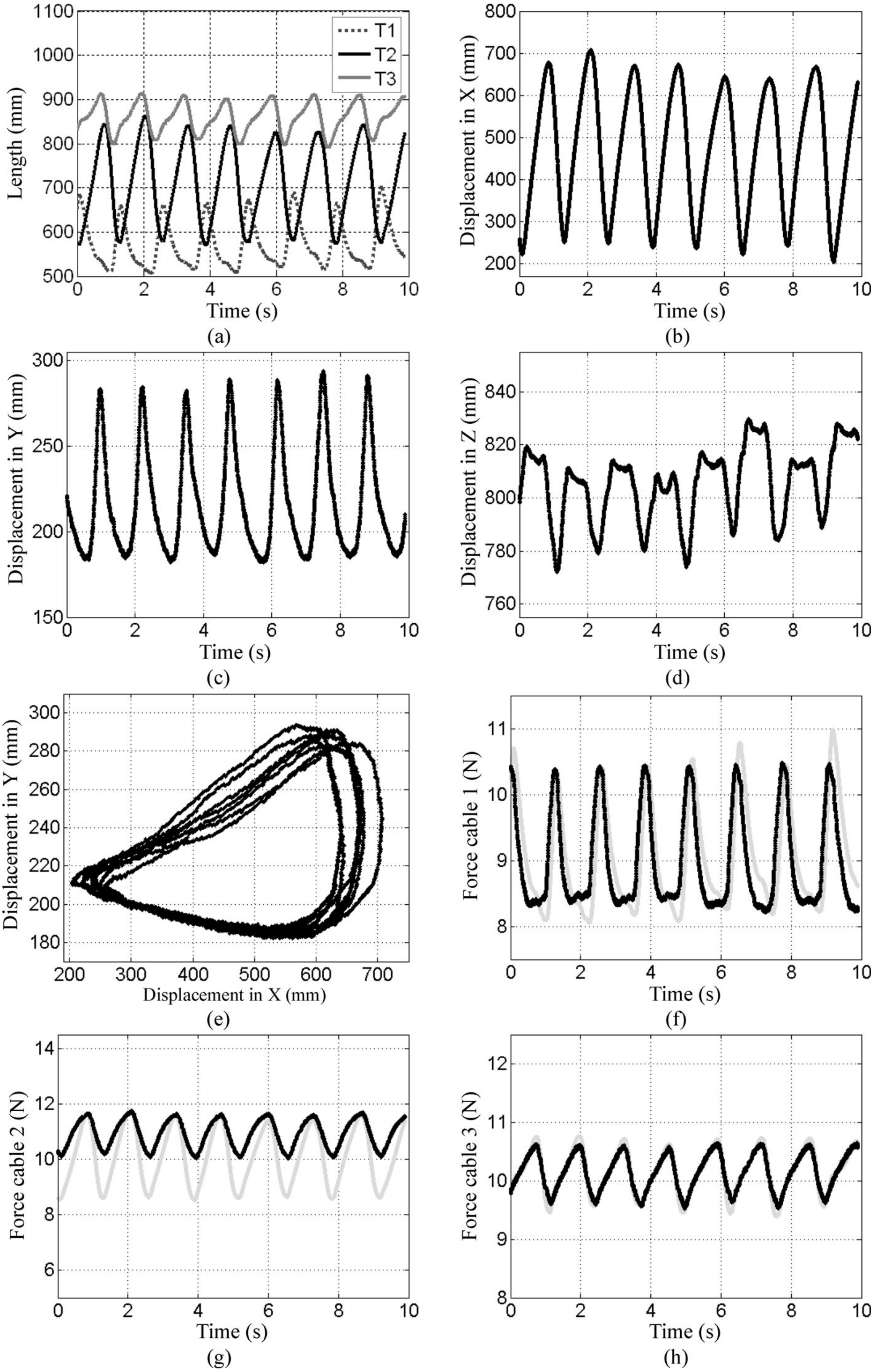

Fig. 17. Results from numerical simulation (in grey lines) and experimental tests (in black lines) for kinetostatic analysis of a human walking: (a) cable lenghts; (b) X motion component; (c) Y motion component; (d) motion component; (e) traiectory in X–Y plane; (f) tension in cable 1; (g) tension in cable 2; (h) tension in cable 3.

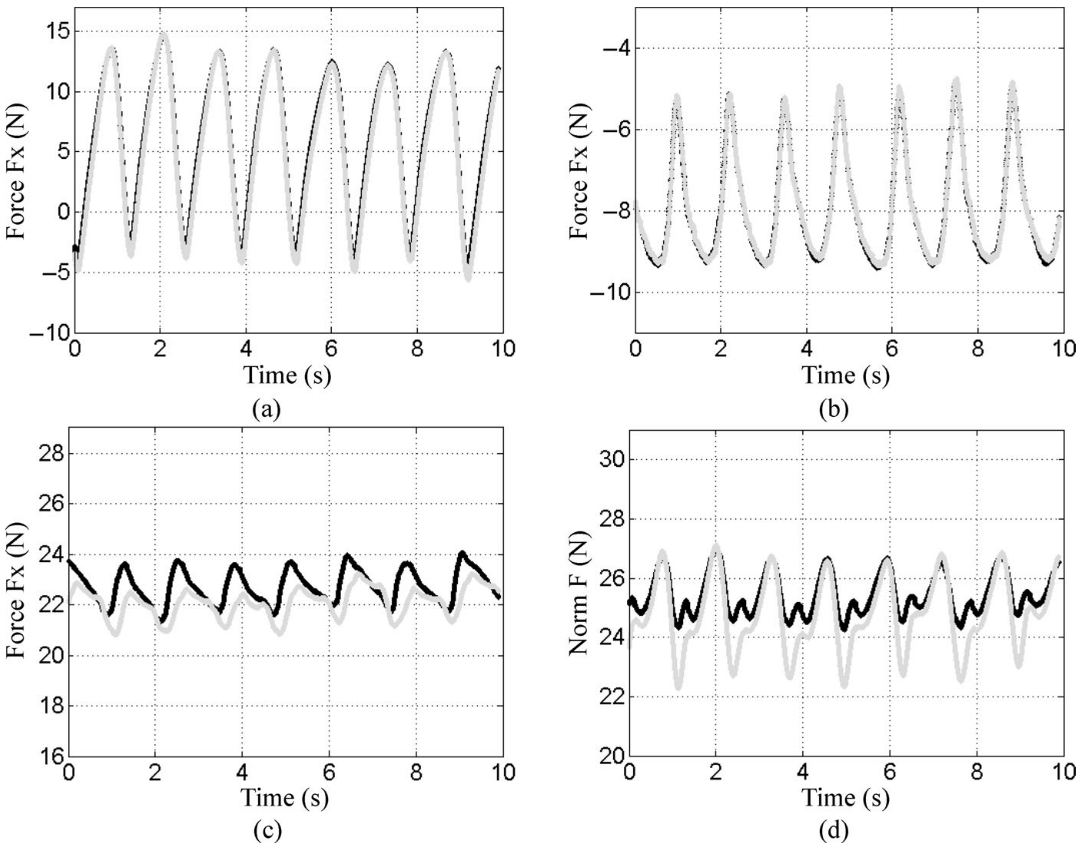

Fig. 18. Results from numerical simulation (in green lines) and experimental tests (in black lines) for kinetostatic analysis of a human walking in terms of forces on the leg end point; (a) X component; (b) Y component; (c) Z component (d) magnitude of force.

movement in the test of Fig. 11(a) and (b), which refers to the hip rotation only in saggital plane. In particular, Fig. 12 shows the displacements and forces of leg end-point along X, Y, and Z axes, and the resulting trajectory and force magnitude.

Figure 13 shows experimental results for the analysis of human leg movement in the test of Fig. 11(c) and (d), which refers to the knee rotation only.

Figure 13 shows the end leg point displacements along X, Y, and Z axes and the corresponding force components together with the resulting trajectory and force magnitude. All experimental results are obtained with a leg end point motion in the X–Y plane. The force in the Z direction is almost constant, as expected.

Figure 14 shows the second type of experiments, which concerns with the human forward and backward motion in the X–Y plane. Figure 15 shows the human forward and backward motion in the Z–Y plane. The experimental results have been obtained by considering two steps forward and two steps backward, as it is shown in Fig. 14 (b) and Fig. 15 (d), which refers to the X and Z displacements respectively, as function of time. It is worth to note that in normal human walking both high and length of a step vary as function of time. The forces that are exerted by the leg during the walking operation have been experienced to be proportional to the leg displacements.

Figure 16 show the experimental set-up for the second type of tests, for which a tapis-roulant has been used to analyze the gait in trasient and steady state phases of walking. In particular, Fig. 16 (d) shows the trajectory of the leg end point for a generic gait. Marker with number 1 is the heel contact (the support phase starts); 2 is the tip contact (the support phase ends); 3 is maximum stride high. The corrisponding configurations are shown in Figs. 16 (a)–(c). It is worth to note that the support contact trajectory has a slope that is related to the tapis roulant sloped configuration. For the reported experimental result in Fig. 16 the maximum high of the step is 110 mm and the step length is 420 mm. Figure 17 shows numerical and experimental results for the steady state of human walking in terms of leg end point displacements and forces exerted by the leg end point. The experimental test refers to seven steps. In particular, X and Y displacements do not vary substantially in term maximum high and length, but if one considers Fig. 17(e) the trajectory does not remain constant but varies at each stride. The Z displacement takes into account again the slope of the tapis- roulant. Figure 17(f) to (h) show comparison between experimental results on the forces in the cables (black line) and numerical simulation (grey line). It is worth to note that the proposed kinetostatic model for three cables gives satisfactory results.

Figure 18 shows a comparison between experimental (in black line) and numerical results (in grey line) for the X, Y, and Z components and magnitude of the force exerted by the limb during the experiments in Fig. 16.

In the paper, we have reported a limited number of results of significant tests that can nevertheless give enough views of the walking characteristics and CaTraSys monitoring capability.

6. Conclusion

In this paper, an application of a cable-based measuring system is presented for an experimental identification of human walking characteristics that can be useful for diagnosis purposes. A new version is proposed for CaTraSys, which has been developed at LARM in Cassino, in order to be used as pose and wrench measuring device. A numerical simulation for a kinetostatic analysis is proposed to characterize the design and operation of new CaTraSys. Once the kinetostatic model has been validated, it can be used to identify different regions of the device workspace as function of the force applied by/on the end-effector.

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