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► **To cite this version:**

Victor de Leon Gomez, Chiara Barone, Aoustin Yannick, Christine Chevallereau. Study of the walking efficiency of a human with a cane. Romansy, Jun 2018, Rennes, France. hal-01845958

HAL Id: hal-01845958

<https://hal.science/hal-01845958>

Submitted on 20 Jul 2018

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Study of the walking efficiency of a human with a cane

Victor De Leon Gomez, Chiara Barone,
Yannick Aoustin and Christine Chevallereau.

Campus Centrale Nantes 1, rue de la Noë, BP 92101. 44321 Nantes, France,
Laboratoire des Sciences du Numérique
de Nantes, UMR CNRS 6004,
CNRS, École Centrale de Nantes, Université de Nantes, France

Abstract This paper proposes a mathematical model of the walking with canes for an anthropomorphic biped with two identical legs with massless feet, two identical arms, and a torso. The walking is performed in the sagittal plane. The walking gait, which is cyclic, is composed of single support (*SS*) phases and impacts. The period of this cyclic walking is determined by the stride, because the motion of coupled arms is synchronized on this stride. The cane is considered massless. Thus in order to compare the walking with and without massless cane, the dynamic dynamic. Numerical tests show that the magnitude of the ground reaction in the stance foot is less with a massless-cane assistance than without one. Especially, the results prove that it is better to use canes with a handle.

1 Introduction

Walking can become difficult for people due to age, injuries disability or medical reasons such as the consequences of a stroke. Thus several works are devoted to the study and development of assistive devices. Since long time ago, the well-known canes/walking sticks or crutches are used to compensate temporary or even permanently a weakness in the lower limbs. Furthermore these assistive devices can sometime improve the walking activity, especially in the case of cerebral disorders. The canes/walking sticks are the most commonly used type of mobility aid than crutches [13]. We can also observe the emergence of new mechanical devices as exoskeletons. Their usefulness depends on the magnitude of the handicap. The current exoskeletons, nevertheless, have limitations and they do not provide autonomous walking. To avoid falls, the use of these existing support exoskeletons for lower limbs require the use of canes [1]. In future, canes will not probably disappear

among the rehabilitation tools. A stick can be also used by a bipedal robot for walking. In [12] a walking control strategy by using a stick is proposed for the walking of a biped robot on uneven terrain. The main advantage of using a stick is to obtain a more robust walking with a good energy efficiency. In [10] it is shown that a stick helps a small-walking robot to move around stably, by supporting most of its weight, and by avoiding an overpressure on its legs. An experimental intelligent cane is presented in [6] and [7]. This cane is composed of an aluminium stick, which is fixed on an omni-directional mobile base with three wheels. A 6-axis force/torque sensor, on the handle of the stick, allows to estimate the user's walking intention. The users has shoes equipped of force sensors to detect a possible fall, then a control law adapts the velocity of the omni-directional mobile base to prevent this fall. Furthermore this cane is connected to a rescue center. In order to monitor the ambulation of Parkinson patients [11] propose also an instrumented cane. These assistive devices modify the interaction of the human with his environnement, especially with the ground.

The dynamic effects of a cane as assistive device is not very well known. [9], with experimental data from capture motions and force platforms, Ivanov and Formal'skii propose a mathematical model of a human walking in sagittal plane. They show that in crutch walking, the ground reaction applied to the feet are lower than the reaction forces applied to the crutches. The crutch used to support the human use the axilla as fulcrum. The torque applied to the crutches by the shoulders is comparable in magnitude with the torques developed by the lower limbs. Despite these previous remarkable contributions, there is a lake of theoretical study about the cane walking. This is a gap that needs to be filled in order to design new canes to better overcome a disability on the lower limbs. In this paper we consider a walking of a human with two identical legs with massless feet, a torso, and two arms. Due to the motion of the arms, which is different from one step to the next step, the walking gait is determined over two steps. The stride is the period of the walking gait. Both steps of the stride are respectively composed of a *SS* phase and an impact. The assistance of the canes is defined by optimizing the torques, which are provided by the human. We compare the efficiency of the assistive canes, which takes place during the first step, and the redistribution of torques and ground reaction by assuming a weakness on the stance leg. For the assistance of the canes we consider two cases: 1) when the person applies only a force on the handle of the cane; and 2) when the person applies a force and a moment on this handle.

This paper is outlined as follows. The dynamic model of the human with and without canes is defined Section 2. The walking reference trajectory is presented Section 3. In Section 4, the statement of problem is developed.

The numerical results are shown in Section 5. Section 6 offers our conclusion and perspectives.

2 Dynamic model of the human with and without walking canes

The planar geometric model of a human without and with walking canes is shown in Fig. 1. This model is composed of two identical legs with a knee and a massless foot, a torso, two identical upper arms and two identical forearms linked in the sagittal plane by revolute joints at the hip, shoulders, ankles and elbows. The walking canes are also assumed massless. Although there are two upper arms, two forearms and two walking canes in this model, the arm segments and the canes will be named in singular throughout the rest of the text because the motion of both arms are assumed identical and coupled. The parameters of the human model are taken from the literature [8]. The whole mass of the biped is 75 kg , its height is 1.75 m . Table 1 gathers the parameters of the model of human. The head of the human is taken into account in the parameters of the torso. Considering the human

	Mass (kg)	Length (m)	Inertia moment ($\text{kg}\cdot\text{m}^2$)	center of mass (m)
Human calf	$m_s = 4.6$	$l_s = 0.55$	$I_s = 0.0521$	$s_s = 0.324$
Human thigh	$m_t = 8.6$	$l_t = 0.45$	$I_t = 0.75$	$s_t = 0.18$
Human torso	$m_T = 41$	$l_t = 0.75$	$I_T = 11.3$	$s_T = 0.386$
Human upper arm	$m_{ua} = 3.9$	$l_{ua} = 0.35$	$I_{ua} = 0.044$	$s_{ua} = 0.16$
Human forearm	$m_{fa} = 4.28$	$l_{fa} = 0.45$	$I_{fa} = 0.05$	$s_{fa} = 0.22$
Massless feet	–	$l_p = 0.30$	–	–
Cane	–	$l_c = 0.8$	–	–

Table 1. Physical parameters of the human subject.

in *SS* phase with a flat foot of leg 1 and that his feet are massless, its configuration can be described with the following generalized vector:

$$\mathbf{q} = [q_1, q_2, q_3, q_4, q_5, q_6, q_7, x, y]^\top. \quad (1)$$

The purpose of the vector definition (1) is to create a dynamic model of the robot. When the stance foot is flat on the ground, an unilateral constraint between the ground and the stance foot is written. When the cane is in contact with the ground, an external force \mathbf{r}_c is exerted by the ground to the tip of the walking cane. As the walking stick is massless, this force is

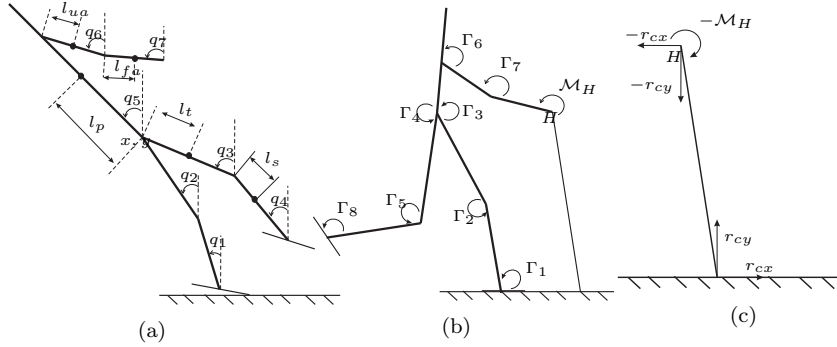


Figure 1. (a) The geometric model, generalized coordinates, and positions of the centers of mass of the links. (b) Inter-link torques of human and the moment \mathcal{M}_H . (c) The forces and moment applied to the walking cane.

the exact opposite of the exerted force from the arm tip to the walking cane Fig. 1 (c). Let r_{cx} and r_{cy} be the horizontal and vertical components of this force \mathbf{r}_c and let \mathcal{M}_H be the moment applied at the point H from the massless walking stick, see Fig. 1 (b). As shown in Figure 1 (b), we show the eight torques applied at the hip-, knee-, elbow-, shoulder-, and ankle-joints. The dynamic model is defined as follows:

$$\mathbf{A}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{h}(\mathbf{q}, \dot{\mathbf{q}}) = \mathbf{D}\mathbf{\Gamma} + \mathbf{J}_{\mathbf{r}_1}^\top \mathbf{r}_1 + \mathbf{J}_{\mathbf{r}_2}^\top \mathbf{r}_2 + \mathbf{J}_c^\top \begin{bmatrix} r_{cx} \\ r_{cy} \\ \mathcal{M}_H \end{bmatrix}, \quad (2)$$

Here $\mathbf{A}(\mathbf{q})$ is 9×9 symmetric positive definite inertia matrix, $\mathbf{h}(\mathbf{q}, \dot{\mathbf{q}})$ is 9×1 vector, which groups the centrifugal, Coriolis, and gravity forces. $\mathbf{\Gamma}$ is 8×1 vector of the seven joint torques applied by the human. Vectors \mathbf{r}_i , with $i = 1, 2$, are the wrench ground reactions applied to the massless feet (and consequently to the ankle-joints). Vector $\mathbf{r}_i = (r_{ix}, r_{iy})^\top$ is composed of the horizontal r_{ix} and vertical r_{iy} components of the ground reaction. Let us remark that the feet being massless we can consider a balance equation between the actuator torque applied at the ankle of the stance leg and the moment due to the ground reaction. Thus considering this balance equation we are able to constraint the center of pressure to stay inside the surface of contact of the sole with the ground for the design walking.

The following constraint equations are correct when the front or/and rear leg is/are on the bearing surface.

$$\mathbf{J}_{\mathbf{r}_i} \ddot{\mathbf{x}} + \dot{\mathbf{J}}_{\mathbf{r}_i} \dot{\mathbf{x}} = \mathbf{0}, \quad \text{for } i = 1 \text{ or/and } 2. \quad (3)$$

The matrices \mathbf{D} , \mathbf{J}_{r_1} , \mathbf{J}_{r_2} , and \mathbf{J}_c can be calculated by using the principle of virtual work (an analysis of matrix D can be found in [5]).

3 The reference periodic walking gait

The goal of this study is to compare the energy consumption during a gait cycle (*i.e.* a stride that is composed of a SS phase on the first leg an instantaneous double support and a SS phase on the other leg and an instantaneous second double), performed by the biped without and with its walking cane. The cane will assist the human only during the first phase. During the second SS phase the arm will transfer the cane from back to front.

The reference walking gait is defined for the biped without the help of the assistive cane. That means that in Section 2 the component of the wrench applied on the cane r_{cx} , r_{cy} , and \mathcal{M}_H are equal to zero. This walking gait is periodic, *i.e.* all strides are identical. The design of the walking over a stride allows to recover the initial position of the massless cane and the arms for performing the next stride. In SS the stance foot has a flat contact with the ground and the torque applied in the ankle of the swing foot is null since massless feet are considered. When the swing foot impacts the ground the other foot takes off. The evolution of each joint of the locomotor system of the biped is prescribed with a polynomial function, of which the coefficients are calculated through a parametric optimization algorithm. The minimized criterion is the integral of the square of joint torques, called *sthenic criterion* [4] for a given walking step length at several velocities; see [2] and [3] for further details. The cyclic walking gait is defined such that the conditions of the flat contact for the stance foot are satisfied, *i.e.* there is no rotation of this stance foot. The maximal magnitude of the joint torques are chosen to be 180 $N.m$. The reference trajectory of the hand is given during the stride, following the motion of a virtual cane Fig. 2. By using the inverse and kinematic models the joint evolution of the elbow and the shoulder are deduced. A sequence of figures that exhibit the stride for this reference trajectory is shown in Fig. 2.

4 Study of the optimal distribution of the torques for the biped with walking cane

The biped with the help of the walking cane tracks the walking periodic gait defined in the previous Section. The cane assists the human only during the first SS phase on leg 1. For the human we assume a weakness in the hip joint of the stance leg during the SS phase. During the second SS the

human brings his cane to the front. During the first SS , since the foot of the swing leg is massless we have $\Gamma_8 = 0$, and the dynamic model (2) includes nine equations. If the human applies a force and an moment on the handle of the cane there are 12 following unknown variables which are:

$$\Gamma_1, \Gamma_2, \Gamma_3, \Gamma_4, \Gamma_5, \Gamma_6, \Gamma_7, r_{1x}, r_{2y}, r_{cx}, r_{cy}, \mathcal{M}_H. \quad (4)$$

Thus, an optimization problem at each sampling time of the first support phase is defined by choosing variables r_{cx} , r_{cy} , and \mathcal{M}_H to track the walking periodic gait. The dynamic model (2) can be rewritten:

$$[\mathbf{D}_1, \mathbf{D}_2, \mathbf{D}_3, \mathbf{D}_4, \mathbf{D}_5, \mathbf{D}_6, \mathbf{D}_7, \mathbf{J}_{\mathbf{r}_1}^\top] \begin{bmatrix} \Gamma_1 \\ \vdots \\ \Gamma_7 \\ \mathbf{r}_1 \end{bmatrix} = \mathbf{A}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{h}(\mathbf{q}, \dot{\mathbf{q}}) - \mathbf{J}_c^\top \begin{bmatrix} r_{cx} \\ r_{cy} \\ \mathcal{M}_H \end{bmatrix}, \quad (5)$$

where \mathbf{D}_i with $i = 1 \dots 7$ is the i -column of matrix D .

If the human applies a force only on the handle of the cane there are 11 unknown variables which are:

$$\Gamma_1, \Gamma_2, \Gamma_3, \Gamma_4, \Gamma_5, \Gamma_6, \Gamma_7, r_{1x}, r_{2y}, r_{cx}, r_{cy}. \quad (6)$$

In this case the optimization problem deals with two chosen optimization variables r_{cx} , r_{cy} to track the walking periodic gait. The dynamic model (2) can be rewritten:

$$[\mathbf{D}_1, \mathbf{D}_2, \mathbf{D}_3, \mathbf{D}_4, \mathbf{D}_5, \mathbf{D}_6, \mathbf{D}_7, \mathbf{J}_{\mathbf{r}_1}^\top] \begin{bmatrix} \Gamma_1 \\ \vdots \\ \Gamma_7 \\ \mathbf{r}_1 \end{bmatrix} = \mathbf{A}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{h}(\mathbf{q}, \dot{\mathbf{q}}) - [\mathbf{J}_{c_1}^\top, \mathbf{J}_{c_2}^\top] \begin{bmatrix} r_{cx} \\ r_{cy} \end{bmatrix}, \quad (7)$$

where $\mathbf{J}_{c_1}^\top$ and $\mathbf{J}_{c_2}^\top$ are the two first column of the matrix \mathbf{J}_c^\top .

For people with disabilities, the minimisation of the human action for walking is an obvious criterion. Thus the following criterion for the first step will be considered.

$$C = \mathbf{\Gamma}^\top \mathbf{\Gamma}. \quad (8)$$

Constraints are defined to ensure that the vertical component of the ground reactions both on the stance foot and the cane tip is positive. Furthermore, a constraint on the centre of pressure is applied in order to keep it inside the surface the sole of the stance massless foot.

In conclusion the statement of the problem is such that, let us define a given walking trajectory with its intrinsic features, how a human with his upper limbs and an assistive cane can compensate a weakness of his stance leg, what is the new distribution of the efforts between the lower and upper limbs.

5 Numerical results and discussion

Figures 3-4 are devoted to the three cases, which are respectively: the human walking without a cane (blue curves); the human walking using the cane with an effort wrench with the components r_{cx} , r_{cy} , and \mathcal{M}_H (red curves); and the human walking using the cane by an effort force with the components r_{cx} , r_{cy} only (black curves). For the assistance with the cane, the magnitude of the applied forces along x -axis and y -axis are limited to 300 N. Of course it is physically possible to consider a greater limit magnitude but we would like to focus our study for elderly people with limited muscular capacities of the upper limb. As shown in Fig. 3 (a) the influence of the cane is big, as expected, in the ankle joint. That means that the magnitude of the ground reaction is much less with the cane as shown in Fig. 4 (a). We can also observe this effect in the knee and the hip of the stance leg. Moreover, there is also an influence of the cane on the swing leg. Fig. 4(b) shows the wrench applied by the cane on the hand for both cases. When the whole wrench is considered we can observe a big reduction of the joint torques for the knee and the hip of the stance leg, unlike when only the force is considered. However, there is no difference in the reduction of joint torques for the swing leg for the two cases of assistance. This simulated "handle" does not influence the vertical component of the ground reaction in the stance foot of the biped, Fig. 4 (a). There is just a small difference at the beginning for the vertical component of the ground reaction. The horizontal component of the ground reaction the difference is more influenced by the simulated "handle". We can observe that in general, a cane with a handle that can deal with the whole wrench is better for the assistance of the stance leg than a cane with a handle that deals only with the force.

6 Conclusion and perspectives

This paper is devoted to a preliminary study about the walking with a cane. This assistive device is well-known from long-time ago but few studies are devoted to its bio-mechanical benefits and thus the possible improvements of its geometric design. With a simple planar model we can analyze nu-

merically the main features, which are experimentally well-known, of the walking cane. We proved that it is better to have a cane with a handle. In the future, it will be interesting to investigate the optimal values of the initial and final position and orientation of the cane. It will be interesting to look for the best physical parameter of the cane, its length, shape designed specifically for one human.

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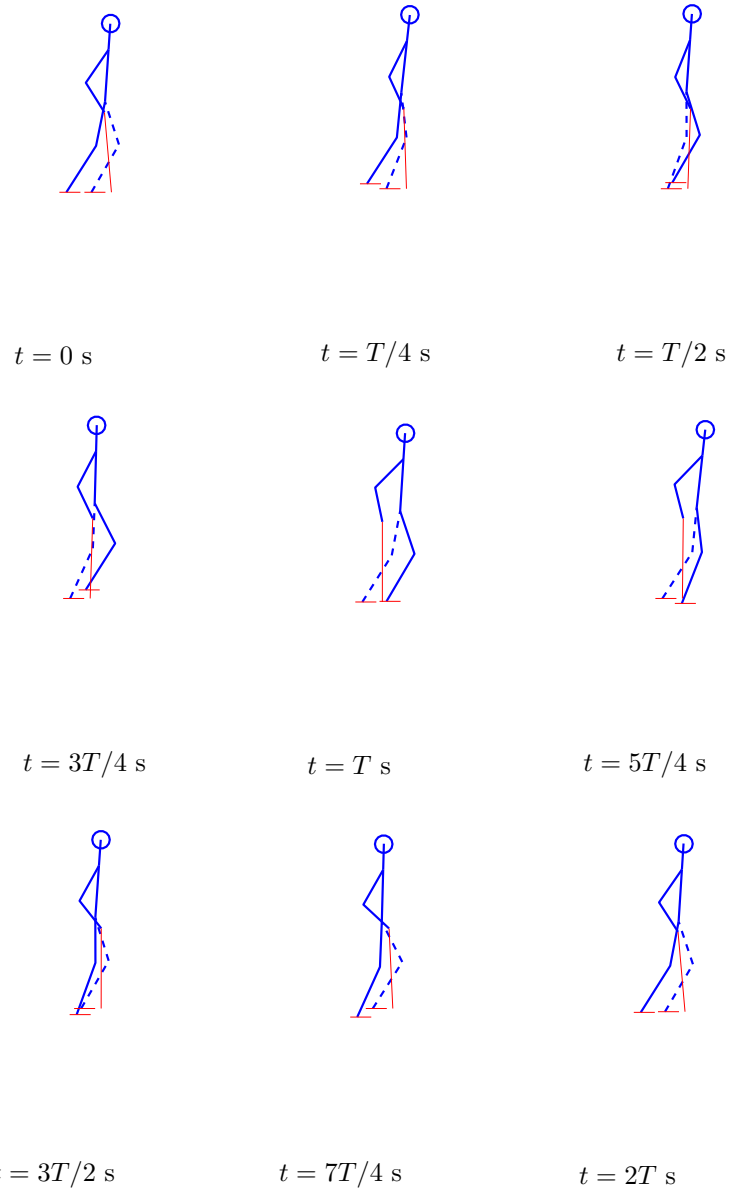
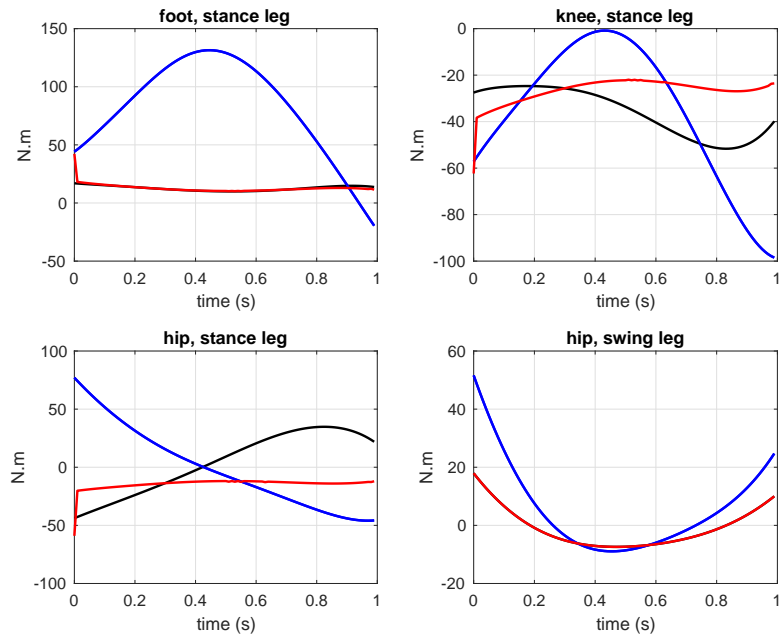
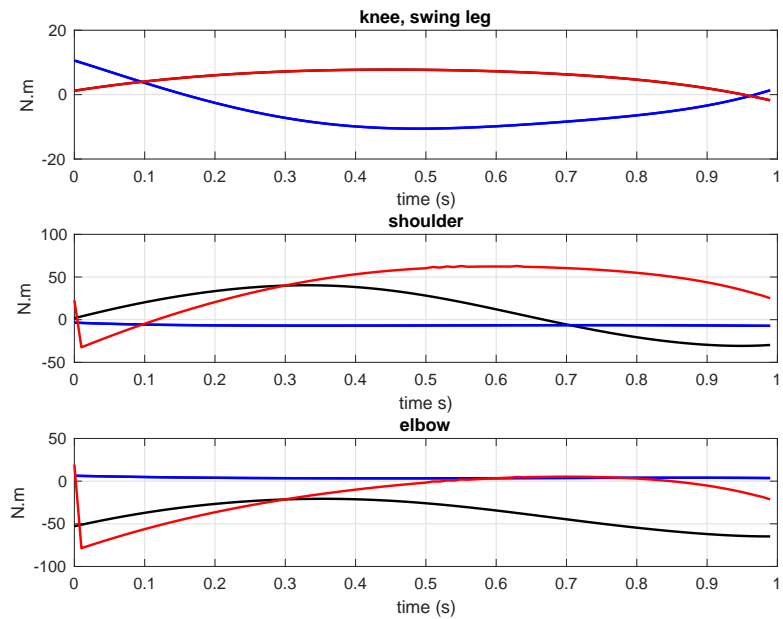


Figure 2. Biped alone, walking gait as a sequence of stick figures.

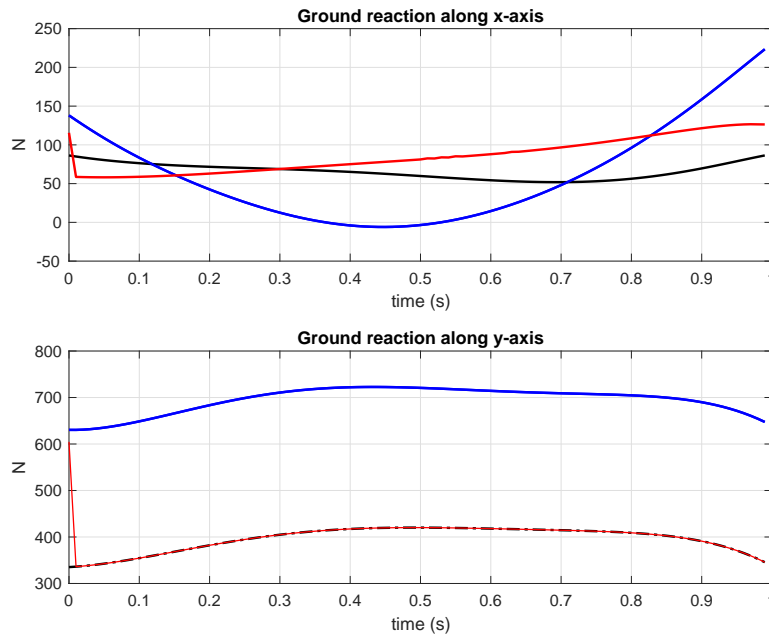


(a)

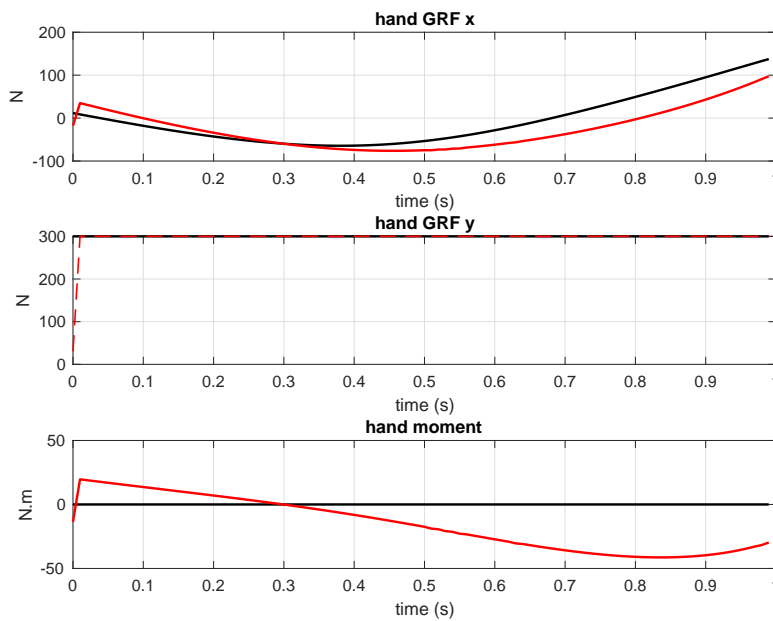


(b)

Figure 3. Profiles of the torques applied in the seven joints, without assistance "blue curves"; by considering a wrench interaction between the human and the cane "red curves"; and by considering only force interaction between the human and the cane "black curves".



(a)



(b)

Figure 4. (a) Profiles of the ground reaction components in the stance foot without assistance "blue curves"; by considering a wrench interaction between the human and the cane "red curves"; and by only considering a force interaction between the human and the cane "black curves". (b) Profiles of the efforts applied by the human to the cane: by considering only "black curves", by considering the wrench "red curve".