# DESIGN OF A SENSORIAL AND DRIVING LOCOMOTION INTERFACE

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Abstract : This paper deals with the design of a 1D locomotion interface of which main characteristic is to be a sensorial and driving force interface. The aim is to keep globally the user at the same place according to forces he/she applied on the interface. So, our interface is instrumented with force sensors and tracking devices. Motor sizing hardly depends on kinematics and dynamics features of human gait. To ensure the computation in real time, tasks of the application are distributed using parallel computing.


Keywords : Walking, systems design, kinematics, tracking systems, parallel computation.

## 1. INTRODUCTION

Many locomotion interfaces have been designed since the soaring of Virtual Reality. A locomotion interface restricts motion to a confined space, but allows unrestrained roaming in the virtual environment by tracking user's limbs or resultant body motion. These interfaces can be divided into three main categories according to their mechanical structure. Firstly, pedalling devices such as *Sarcos Uniport* (Hollerbach, 2002) are composed with a stationary bicycle. Contrary to pedalling devices, 1D or 2D treadmill devices allow the user to walk more naturally. For instance, the *Torus Treadmill* (Iwata, 1999) enables the user to walk within a plane area. Finally, programmable foot platforms such as *GaitMaster* (Iwata, 2001) propose to support the user with two pedals actuated by six degrees of freedom. More recently, a new locomotion interface called *CirculaFloor* (Iwata, *et al.*, 2004) has been designed using a group of movable tiles. The movable tiles employ holonomic mechanism which performs omni-directional motion.

Considering above description, our contribution presented in the paper is the design of a locomotion interface which simulates natural walking in one direction while globally keeping the user at the same place. Contrary to interfaces introduced before, our interface is a sensorial and driving force interface. That is to say, our interface is composed with actuators to keep, on the whole, the user at the same place according to forces applied by the user on the active ground. This functioning is closed to that of haptics. As a consequence, the design of our 1D locomotion interface will allow natural walking in place by controlling each foot independently. Intended applications of our interface are numerous.

It can help doctors to analyse aberrant gaits by comparing the gaits of patients before and after treatment to historical case studies of normal and abnormal gaits. It could also be useful for health rehabilitation by imposing a pre-recorded motion sequence to move one of the user's feet for hemiplegic patient. In addition to that, our interface would be suitable for whole body exercise. Finally, it can be used to move virtually in immersive environments.

First, this paper aims at describing specifications of our locomotion interface, whereas the next part is devoted to its design. Finally, the global architecture of our interface will be described.

## 2. SPECIFICATIONS

Our purpose is to design a locomotion interface which enables the user to walk naturally in one direction while his position is maintained. In addition to that, each foot has to be controlled independently at any time. To overcome this problem, the design of our interface is based on pedalling devices, where each pedal has 1 degree of freedom in translation. First, the meaning of natural walking has to be defined. The gait cycle begins when one foot contacts the ground and ends when that foot contacts the ground again.

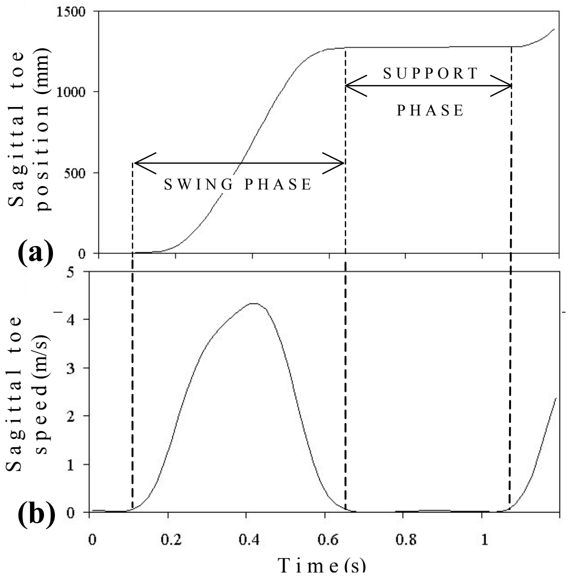


Fig. 1. Position and speed of foot toe in sagittal plane

Thus, each cycle begins at initial contact with a stance phase and proceeds through a swing phase until the cycle ends with the limb's next initial contact. Stance phase accounts for approximately 60 percent, and swing phase for about 40 percent of a single gait cycle. Each gait cycle includes two periods when both feet are on the ground. The first period of double limb support begins at initial contact, and lasts 10 to 12 percent of the cycle. The second period of double limb support occurs in the final 10 to 12 percent of stance phase. As the stance limb prepares to leave the ground, the opposite limb contacts the ground and supports the body's weight. Both periods of double limb support account for 20 to 24 percent of the gait cycle's total duration. The kinematic features of human walking (Multon*, et al.*, 1999) onto a plane ground can be obtained thanks to a 3D optical motion capture device called VICON. It gives us the position of markers placed on user's foot during normal walking. The trajectory and speed of foot toe in the sagittal plane during walking in a straight line are presented in Figure 1. It appears the average step length is about 1.2 m. Moreover, foot speed has a parabolic shape and can be up to 4.5 m/s, whereas the acceleration can be up to 28 m/s². The average speed is 1.3 m/s corresponding to the speed of the user's center of mass.

So, both phases of gait cycle have to be taken into account to design our interface : the stance phase and the swing phase. During the stance phase it is necessary to measure the ground reaction forces in order to compute the movement of the pedal while keeping the user at the same place. For our 1D interface, the forces applied on the sagittal plane are enough to compute the trajectory of pulling back. Furthermore, user's feet has to be tracked during the swing phase to ensure, for safety reasons, the pedal to be always under the foot. Consequently, the pedal has to be instrumented to satisfy all these requirements.

## 3. LOCOMOTION INTERFACE DESIGN

### *3.1 Hardware framework of the locomotion interface*

This part presents the design of our 1D locomotion interface. First, a free motion is let to the foot which is in the swing phase, that is to say when there is no contact with the ground surface. But, the interface will act on user's feet during single and double stance phase in order to keep him at the same place. To satisfy all these requirements a cartesian structure has been chosen. That is to say our 1D locomotion interface will be composed with two linear independent axles, 2 meters long, each one having one pedal.

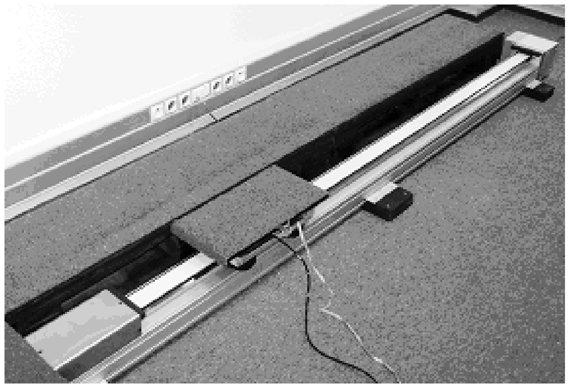

Fig. 2. Linear drive unit

This paper presents the implementation carried out for one of both linear axles. The main advantage of our device is its modular structure. Indeed, it is easy to upgrade the actuation mechanism to add new mechanism for 2D or 3D locomotion. Therefore, our 1D locomotion interface is composed with two linear axles. Each axle is a belt driven linear transmission such as illustrated in Figure 2, which is built on a compact aluminium beam fitted with V slides. The drive is provided by a driving belt and pulley to give rigidity, speed and accuracy. Moreover, each axle is fitted with a geared brushless motor in order to fulfil the kinematic and dynamic requirements of walking.

During the swing phase, it is necessary, for safety reasons, to track user's foot so that the pedal is always under the foot. That is the reason why brushless motors are suitable for our application because they can provide high speed and dynamics. This motor is equipped with a resolver that gives the rotor position. Concerning the control of the brushless motor, the ST7MC microcontroller can run dedicated and special control laws. Evaluation of brushless motor power is described below.

<u>Calculation of motor speed and torque</u>. Brushless motor drives each linear axle. So as to evaluate the right speed and torque needed for our locomotion interface, it is first necessary to define the duty cycle. Indeed, the duty cycle is composed with two phases : the first one corresponds to the swing phase when the pedal has to track user's foot whereas during the second phase, the pedal is supporting and pulling back user's foot. Let X be the direction of walking and Z, the vertical axis. Therefore, the most critical phase is the second one. Indeed, during this phase, efforts in X and Z axis are applied on the pedal depending on the global position of the user. Obviously, these efforts have to be taken into account by applying the fundamental principle of dynamics to the pedal, such as presented just below in equation 1.

$$M_p \frac{d^2 x_p}{dt^2} = F_x + F_M + (F_z + M_p g).\tan\varphi \quad (1)$$

Where :

- $x_p$ is the horizontal position of pedal,
- $M_p$ is the pedal mass,
- $F_x$ and $F_z$ are the forces applied by the user on the pedal,
- $\tan\varphi$ is the guides friction coefficient,
- $F_M$ is the force generated by the motor.

So, by imposing a speed trajectory to the pedal during the stance phase, it is easy to define the horizontal force which must be applied on it to suit the trajectory. Considering this, equation 2 defines the motor torque corresponding to this horizontal traction force.

$$C_m = \left( J_m + \frac{J_P}{\eta_R.K^2} + \frac{M_p.R_p^2}{\eta_R.\eta_C.K^2} \right) . \frac{d\Omega_m}{dt} - \frac{\left(F_X + (F_Z + M_p g).\tan\varphi\right).R_p}{\eta_R.\eta_C.K} \quad (2)$$

Where :

- $C_m$ is the motor torque,
- $J_m$ is the motor inertia,
- $J_p$ is the pulley inertia,
- $K$ is the gear reducer ratio,
- $\eta_c$ is the output of linear transmission,
- $\eta_r$ is the output of gear reducer,
- $R_p$ is the pulley radius
- $\Omega_m$ is the motor angular speed.

On the other hand, evaluation of motor speed is required during the duty cycle. During the swing phase, the pedal is tracking user's foot and has to be always under it. Regarding Figure 1, it has been decided to fix the maximum horizontal speed of the pedal equal to 3 m/s which corresponds to 3000 rpm for the brushless motor with *K* equal to 5.

<u>Simulation part</u>. In order to compute the precise torque needed for our application, a simulation has been run with Matlab Simulink software. This simulation computes the motor speed and torque required to enforce the trajectory of user's foot when it is pulled back during the stance phase. It is important to underline the fact that motor speed and torque hardly depend on the trajectory of pulling back. First, a trajectory with a constant speed of 1.5 m/s has been considered. In that case, this speed has already been reached when user's foot first interacts with the pedal so that the acceleration is equal to zero. To do so, it is necessary to accelerate the pedal before the foot interact with the pedal. The simulation parameters concerning the forces applied by the user during walking have been measured thanks to a force platform with a person of about 65 Kg. Obviously, since there is no acceleration, there is no effect of inertia and the simulation shows that the maximum value of motor torque is about 1.5 N.m. However, the pedal should need to be accelerate or decelerate during the stance phase so as to simulate floor interactions. In that case, inertia effects are not insignificant.

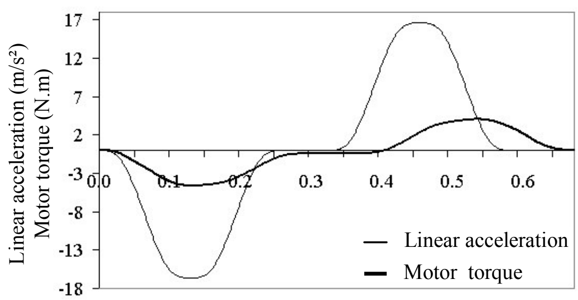


Fig. 3. Simulation of motor torque

For instance, during the stance phase the pedal follows a trajectory whose acceleration is presented on Figure 3. The acceleration and deceleration of the pedal are up to about 17 m/s². In this case, motor torque computed appears to be more important than in the case where there is no acceleration. Indeed, Figure 3 indicates that motor torque should be up to 4.5 N.m. If the motor speed necessary to enforce this trajectory is also taken into account, it is easy to estimate the motor power needed at any time of the duty cycle. Here, the maximum motor power is near to 600 W. So, it has been decided to use a brushless motor which can provide a continuous torque of 4 N.m with a rotation speed of 3000 rpm.

### *3.2 Instrumentation of the pedals*

Our locomotion interface is instrumented in order to track user's feet during the swing phase and to measure efforts applied by the user on the pedals. Indeed, it is important to keep in mind that our interface is a sensorial and driving interface : the interface will act on the user in relation with the efforts measured during walking. Therefore, it is first necessary to develop a tracking system of the foot during the swing phase. Electomagnetic tracking devices are very efficient but the main drawback is that they cannot be used with our interface because of the magnetic field created by the motor. Ultrasonic tracking devices do not have the disadvantage of electromagnetic tracking devices but their accuracy is too low for our application.

Thus, it has been decided to design a 2D mechanical tracking device which is based on cable and pulley equipped with a drum. The linear movement of a 1 m long cable is converted into a rotational movement by a measuring drum. The measuring drum is coupled to the shaft of the encoder. A displacement of the cable tip corresponds to a change in angle on the encoder shaft. In order to leave user's foot free during the swing phase, the extraction force of the cable is negligible. The tracking device placed on the back of the pedal has been designed. The extremity of the cable is attached to the heels of user's feet. The restoring force of a spiral spring holds the measuring cable tight and prevents errors due to slackness. Accuracy of the whole system is 0.1 mm for 1024 counts per turn of the encoder and the maximum cable speed is 10 m/s whereas the maximum acceleration is 100 m/s². So, our tracking device fulfil kinematics and dynamics requirements for tracking human feet during the swing phase of walking.

In order to control the pedal to be always under user's foot during the swing phase, it needs to be precise the position of the foot in the direction of walking. During the swing phase, it is supposed that user's foot has a trajectory on the plane defined by X and Z axis since displacements in Y direction are insignificant. So, it is necessary to have at least two mechanical tracking devices to track one foot, each one giving a distance between the foot and the pedal. To overcome this problem, a low cost method is proposed using only one mechanical tracking device to track one foot.

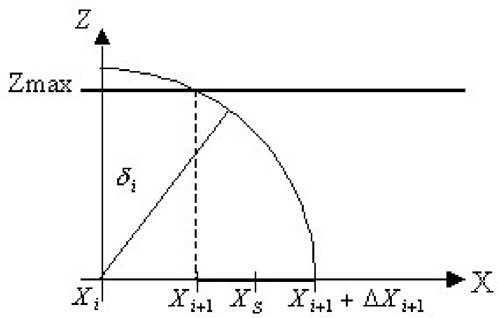


Fig. 4 . Approximation of user's foot position

Where :

- $X_i$ is the position of the pedal at time $t_i$ .
- $Xs$ is the estimation of the foot new position.
- $\delta_i$ is the distance between foot and pedal.

As shown in Figure 4, a value of Zmax is chosen symbolising the maximum height of foot during the swing phase. The aim is to estimate the position of the foot $X_{i+1}$ thanks to the position $X_i$ and $\delta_i$ . In the case $\delta_i$ >Zmax, the position of the foot is in the interval $[X_{i+1}, X_{i+1} + \Delta X_{i+1}]$ and the position of the foot $Xs$ is evaluated as the middle of this interval. Therefore, the evaluated position of user's foot is given by equation 3 :

$$X_S = X_i + \frac{\delta_i}{2}.\left[1 + \sqrt{1 - \left(\frac{Z\max}{\delta_i}\right)^2}\right] \quad (3)$$

When $\delta_i$ <Zmax, it is decided not to translate the pedal, that is to say $Xs = X_i$ . In that case, the accuracy of the system is Zmax. An improvement to this algorithm is to use a non-constant value of Zmax depending of the time. Indeed, during normal gait, the values of Zmax are evolving with a parabolic shape whose maximum depends of user height and step length. This algorithm has been tested with a natural walking and gives an accuracy of 15 mm for each cycle of walking. It is also important to underline the fact that the positioning error does not cumulate at every cycle. Nevertheless, this algorithm do not prevent the user from walking abnormally. Indeed, if he/she raises his/her foot too high in the direction of Z axis, the pedal will go falsely ahead. Consequently, an other degree of freedom in rotation should be added around Y axis in order to measure the slope between the cable and the X axis. As a consequence, the position of user's foot would be defined thanks to its polar coordinates.

On the other hand, the ground reaction forces applied by the user need to be known on each pedal in order to compute the pedal trajectory when it is pulling back user's foot. Furthermore, the measurement of ground reaction forces are necessary for human gait analysis. Here is presented the design of an 1D force platform with the pedal.

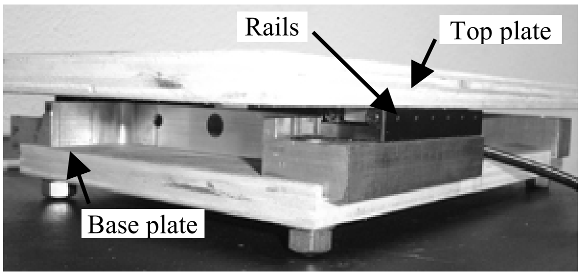


Fig.5 . 1D force platform

Our force platform consists of a top plate connected to a base plate separated by a force sensor and three rails. All sagittal loads applied to the top plate must go through this sensing element in order to be measured correctly. The sensing element is a S-shape sensor which works in traction/compression mode. One tip of the sensor is mounted on the base plate whereas the other one is fixed to the top plate. The transmission of sagittal forces is ensured by the three rails which guide the top plate such as indicated on Figure 5. In our case, walking produces dynamic loading so it is necessary to mount correctly the force platform to the pedal so as to minimise vibrations.

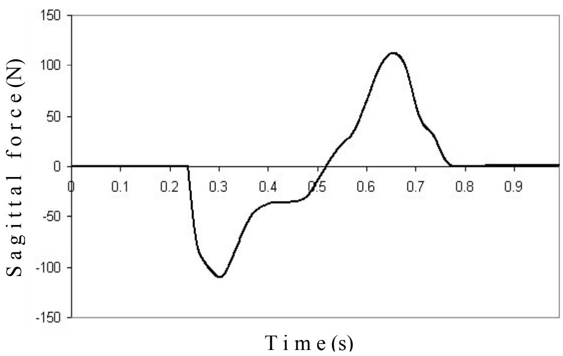


Fig.6 . Sagittal forces measurement

Our low-cost 1D force platform has already been tested. Figure 6 shows the sagittal forces applied by the user on the pedal during normal gait. Results are similar to those reported in scientific papers about biomechanics.

### *3.3 Locomotion interface control*

The foot contact detection task is necessary to know if the application is running in swing or stance phase (switch function). Figure 7 presents a hybrid model of motor control computation. On the one hand, during the swing phase, the tracking algorithm uses the measurements of the distance between the foot and the pedal to compute the motor control. The implemented control law can be either a PID or a RST controller. On the other hand, during the stance phase, the interface can be compared to a particular haptic device (Anthierens, *et al.*, 2004) which compute a displacement according to a force applied on it. So, the force platform measures sagittal forces and computes the trajectory of pulling back thanks to direct dynamic relations.

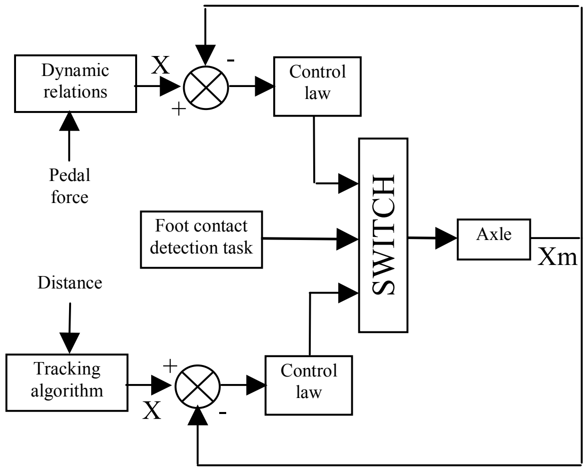


Fig.7 . Functional description

## 4. INTERFACE IMPLEMENTATION

This part aims at describing the implementation of our 1D locomotion interface. In order to acquire all signals required to run our application, a digital acquisition board is used. In order to be as close to real time as possible it has been decided to dedicate one PC to signals acquisition. Moreover, our application needs to compute several tasks, namely the foot contact detection with the pedal to switch between swing and stance phase, the tracking of user's foot during the swing phase and the trajectory computing of pulling back during the stance phase. These tasks will provide the motor control to the ST7MC motor control board. Then, the motor control board will drive the brushless motor. Finally, the control law runs on the motor control board and coefficients of control law can be easily changed in real time to suit the different phases of the cycle.

In order to give the user a visual rendering, a virtual scene has been simulated in which he/she can move according to his/her walking in the real scene. This vision rendering should prevent the user to look at his feet, such as during a natural walking. Furthermore, 3D sound rendering (Yushchenko, *et al.*, 2004) is also a way to drown out the noise generated by the interface. Thus, visual and sound rendering are two important features used to prevent the user from being disturbed by the locomotion interface. To run the virtual scene, an other task is required to compute the virtual eye position thanks to linear drive data. This task is not totally independent because it needs the linear drive data. Output data of this task are needed to perform the independent simulation task.

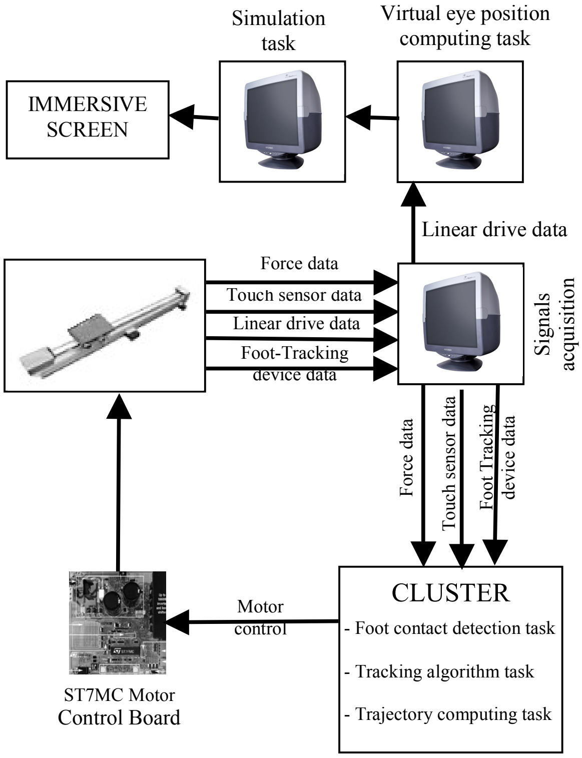


Fig.8 . Global architecture

All these tasks cannot but entail an important amount of numerical computing. Considering the fact that each task is specific, it is more difficult to program one task that would carry out all the tasks presented above rather than programming each task independently. Therefore, all the tasks have to be distributed using parallel computing. As shown in Figure 8, PVM (Parallel Virtual Machine) is used to partition the overall workload in terms of different operations which could have different frequencies. Indeed, in our case all the tasks do not have the same loop frequency. For instance, the virtual eye position computing task has a frequency lower than the tracking one.

Using PVM (Geist, *et al.*, 1994) also enables to perform two different tasks in parallel which is needed for our application. Parallelism is obtained by concurrently executing all programs and by establishing a data "pipeline" between them. For example, the virtual eye position computing task and the tracking algorithm task have to run in parallel, otherwise there would be no synchronisation between the real displacement on the interface and the virtual one. Finally, our function decomposition approach is also justified by the low amount of total exchanged data.

## 5. CONCLUSION AND FUTURE WORKS

This paper presents the design of a 1D locomotion interface respecting specifications related to human gait. Our interface has a cartesian mechanical structure and consists of two linear drives with pedals so as to globally keep the user at the same place by acting independently on each foot. The motor sizing hardly depends on acceleration range during the stance phase because of the inertia of all the moving parts. The pedal is also instrumented with low-cost solutions in order to track user's feet during the swing phase and to measure ground reactions forces during the stance phase. Concerning the implementation of the interface, it was decided to use PVM to partition all the specific tasks. Therefore, these tasks can be run in parallel to ensure synchronisation between the user and the interface.

Future works will be to design a 2D locomotion interface. This interface will be based on the 1D locomotion interface presented in this paper. In other words, mechanical structure and instrumentation part will be improved for walking in a plane.

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