DYNAMICAL BEHAVIOR OF A WHEELCHAIR UNDER THE MOST COMMON DEFORMITY OF THE PROPULSION CYCLE

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Abstract. *The wheelchair is the most accessible solution for mobility problems, but driving the system requires physical capacity and motor coordination, otherwise the system behavior will not respond to users' wishes. It is, therefore, important to know the main deformities that can occur during the propulsion cycle and their effects over the wheelchair displacement. From this, it is possible to develop control strategies to correct these deformities and it becomes easier for the user to drive his wheelchair. The dynamic model used to simulate the wheelchair was a doubled bicycle model with driving force applied over the rear wheels and steering over the front wheels. The main deformities that can occur during the wheelchair propulsion are insufficient propulsion force, sagital disparity and propulsion time delay. All these deformities will result in different displacements if compared with a standard propulsion cycle for forward movement. The results of this article are important to develop control strategies to correct propulsion force deformities and to identify user propulsion problems by the observation of a usual wheelchair displacement. These control strategies can increase users mobility and improve their quality of life.*

Keywords*: model, propulsion force, deformity, wheelchair*

1. Introduction

Wheelchairs have undergone great development in recent years. Studies of their dynamic behavior, however, have not practically occurred. Such studies are important because they allow the improvement of the user mobility.

Recent works by Arva et al. (2000) and Corfman et al. (2000) show the importance of having a more independent life. This contributes for a better development of usual abilities as interpersonal communication, psychosocial skills and other important skills to live in a society.

As stated by Lombardi Jr. and Dedini (2005), Guo et al. (2003), Stein et al. (2003), Wei et al. (2003) manual wheelchair users can develop Dort (damage over repetitive effort), due to the application of repetitive force over the propulsion ring. As this work will show, propulsion force has great influence in the system dynamic behavior.

Small modifications to the propulsion cycle can cause great wheelchair displacement deviations. These deviations make user independent mobility more difficult, because he will need constant path corrections that will increase his physical effort and consequently contribute to a greater risk of developing damages to the muscles.

It is important to be aware of the influences of the propulsion force over the wheelchair dynamic behavior, because it will be useful for the development of control strategies for a servo-assisted wheelchair. The servo-assisted mechanism concept is a group of motors mounted to the wheelchair capable of propelling the system but which will be activated when the user force acting over the propulsion ring is higher than a specific value defined by a doctor or a therapist. (Lombardi Jr. and Dedini, 2005)

A wheelchair servo-assisted mechanism can improve the user locomotion capability and at the same time it does not allow him to remain without making physical exercises. The locomotion improvement occurs because less effort is necessary to propel the wheelchair. (Lombardi Jr. and Dedini, 2005). Furthermore, the control strategies presented in the servo-assisted motorization system can correct the most common propulsion force deformities, making it easier to drive the wheelchair.

The work is divided in three topics. First, the bicycle model will be adopted for the development of the system dynamic equations. Then, the definitions of the propulsion cycle considered standard, according to the literature, will be presented and finally a study of the system dynamic behavior will be carried out under the influence of the sagittal disparity, of the time delay of the propulsion cycle and of both deformities simultaneously.

2. Dynamic Model

The dynamic model is based on the Newton-Euler and Jourdan equations and the aim is to develop the dynamic equations for computational simulation. All equations were written on the non-inertial base over the system CG.

These equations represent the system dynamic behavior related to the propulsion forces acting over the rear wheels, and with a servo-assisted assembly, the motor actuation is added to the propulsion force. It is important to remember that the steering occurs on the front wheels, and the steering forces are always equal to zero because the front wheels are free and the changing direction is made by different forces acting over the rear wheels.

Fig. 1 represents the free body diagram (DFB) of the system wheels; this figure is enough to represent all important system forces, which are the transverse and longitudinal forces acting over the front and rear wheels. Each force is denoted by a sub-index from 1 to 4 which represents the number of the wheels as shown in the figure.

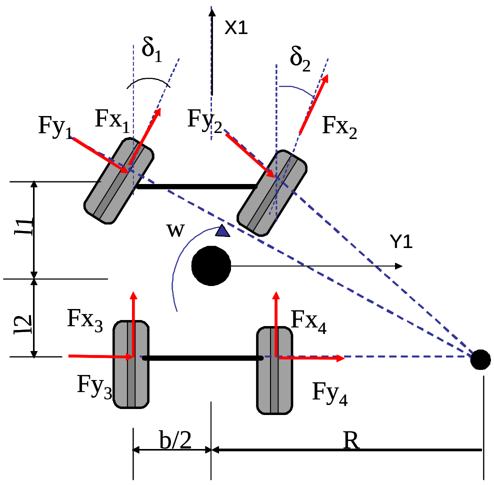


Figure 1. Free Body Diagram for the wheelchair

It can be observed that the longitudinal and the transverse forces are influenced by the resistance forces defined, as presented by Lombardi Jr. and Dedini (2005). Transverse forces are represented in the expression by Eq.(1) for the front wheels 1 and 2 and Eq. (2) for the rear wheels 3 and 4:

$$Fy_i = Cy_f . y_i - \frac{m.g.l2}{2.L}.\cos(j).\sin(g) - Fat_i \tag{1}$$

$$Fy_i = -Cy_t . y_i - \frac{m.g.l1}{2.L}.\cos(j).\sin(g) - Fat_i \tag{2}$$

The previous expressions were developed for a general situation, track (with double inclination), which represents two rotations over the roll axis (X axis, represented by the variable g) and pitch (Y axis, represented by the variable j) angles. The wheels were considered rigid as presented in the work by Lombardi Jr. (2002), so that the lateral force has a component resulting from the multiplication of wheel sliding angle (y) and the rigidity coefficient to the wheel movement (Cy). Variable Fat_i used on previous equations represents the lateral friction force, which is responsible to avoid system lateral slip movement.

For the longitudinal forces the expressions are very similar to the expressions for the resistance forces with the exception of terms *F3* and *F4* in longitudinal forces *Fx3* and *Fx4*, which are the forces applied by the wheelchair user during the manual propulsion or the forces applied by the user and the motorization system during the servo-assisted propulsion. The following expressions, Eq.(3), for the front wheels and Eq.(4) for the rear wheels, represent the longitudinal force for the wheels.

$$Fx_i = -\frac{m.g.l2}{2.L}.\sin(j) - m.\frac{m.g.l2}{2.L}.\cos(j).\cos(b) \tag{3}$$

$$Fx_i = F_i - \frac{m.g.l1}{2.L}.\sin(j) - m.\frac{m.g.l1}{2.L}.\cos(j).\cos(b) \tag{4}$$

The variable m represents the rolling friction coefficient. Now, applying the Newton-Euler and Jourdan equations, it is possible to obtain expressions for $\dot{V}x$, $\dot{V}y$ e $\dot{w}_z$, respectively longitudinal, transverse and angular accelerations of the system. The equations resulting from this procedure are the following:

$$\sum Fx = m.ax \Rightarrow$$
$$Fx_3 + Fx_4 + Fx_1.\cos(d_1) + Fx_2.\cos(d_2) - Fy_1.\sin(d_1) - Fy_2.\sin(d_2) = m.(\dot{Vx} - Vy.w_z) \qquad (5)$$

$$\sum Fy = m.ay \Rightarrow$$
$$Fy_3 + Fy_4 + Fy_1.\cos(d_1) + Fy_2.\cos(d_2) + Fx_1.\sin(d_1) + Fx_2.\sin(d_2) = m.(\dot{Vy} + Vx.w_z) \qquad (6)$$

$$\sum Mz = Iz.\dot{w}_z \Rightarrow$$
$$\frac{b}{2}.(Fx_3 - Fx_4) + l1.(Fx_1.\sin(d_1) + Fx_2.\sin(d_2)) - l2.(Fy_3 + Fy_4) + l1.(Fy_1.\cos(d_1) + \cdots \qquad (7)$$
$$+ Fy_2.\cos(d_2)) + \frac{b}{2}.(-Fx_2.\cos(d_2) + Fx_1\cos(d_1) + Fy_2\sin(d_2) - Fy_1.\sin(d_1)) = Iz.\dot{w}_z$$

Variables d_1 and d_2 represent respectively the steering angle in the front left wheel and front right wheel. It is important to observe that these angles are defined by the propulsion force differences between the rear propulsion wheels. In other words, the system has no independent steering but it is defined by the difference in the propulsion force over the rear wheels.

Using Eq.(5), Eq.(6) and Eq.(7), it is possible to simulate the dynamic behavior for a system comprising a wheelchair and a user, according to the objective of the article. Some graphics are being presented below, showing the dynamic behavior for a manual propulsion wheelchair, over a plane floor. The following simulations are based on the values: *m*, mass of the system, 110 Kg; *L*, distance between the wheelchair axles, 0.5 m; *l1*, distance between the CG (center of gravity) and the front axle 0.3 m; *l2*, distance between the CG and the rear axle, 0.2 m; *b*, wheelchair width, 0.7m; *Izz*, inertia momentum of wheelchair, 6.78 m^4 , *m*, rolling friction coefficient, 0.015; h, CG height, 0.5 m The wheelchair geometric characteristics were based on standard ABNT 9050, and the inertia momentum (*Izz*) for the wheelchair was assumed as being equal to a cube with dimensions *b*x*L*x*h*.

3. Considerations about the propulsion force

In this section only the two main factors for the biomechanical model will be presented. These factors are the propulsion standard and the features of the propulsion force cycle.

For this study, it was considered that movement only happens in the user sagittal plan. This is consistent with the movement observed in recent studies by Souza (2000), Lombardi Jr. (2002). Each manual wheelchair user adapts to a propulsion standard, or, in other words, to a trajectory defined by the upper members during manual propulsion. Souza (2000) reports that there are 4 standard paths for wheelchair propulsion: arc, semi-circle, simple looping and double looping. This paper will use only the arc as upper member movement.

In the same way it will be considered that the propulsion force has regular features and it can be divided in three parts (*T1*,*T2* and *T3*), where *T1* is the time necessary for the user to apply the propulsion force from zero up to his maximum value (*Fmax*), which depends on user characteristics (lesion level, physical condition), *T2* is the time during which force remains constant up to the moment the hand releases the ring and *T3* is the time for the hand to return to its initial contact point to restart the cycle. Values adopted *T3*=1,0 s for plane surface and 0,4 s for slope surface. *T2* was considered 3 times bigger than *T1*.

Combining the above-mentioned two pieces of information, the propulsion force pattern is shown in the following Figure 2:

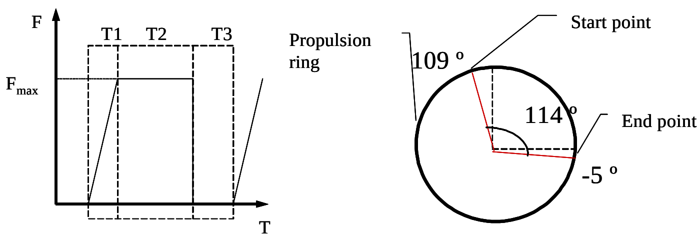


Figure 2. Characteristic Propulsion Force definition

The maximum propulsion force value (*Fmax*) depends on biomechanical characteristics. For this paper, as it aims studying the dynamic behavior of the system, it will be assumed that the user is capable of applying the necessary force for the movement. The necessary force for the movement is the sum of all resistance forces to the rolling movement acting over each wheel and also the weight component contrary to movement, if convenient (slope planes).

Finally in this work the standard propulsion cycle was defined adopting variables *T1* equal to 0.5 s, T2 equal to 1.0 s, and T3 equal to 1.0 s. Furthermore, the propulsion cycle will be repeated 10 times for each simulation. The following figure represents the standard propulsion cycle.

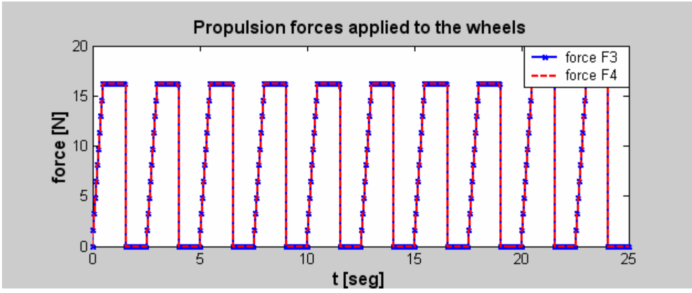


Figure 3. Standard propulsion cycle

4. Manual wheelchair simulation

This section aims at studying the dynamic behavior of a wheelchair under the effects of the most common propulsion force characteristic. This information is important because based on its results it will be possible to quantify the improvement obtained using a servo-assisted motorization system.

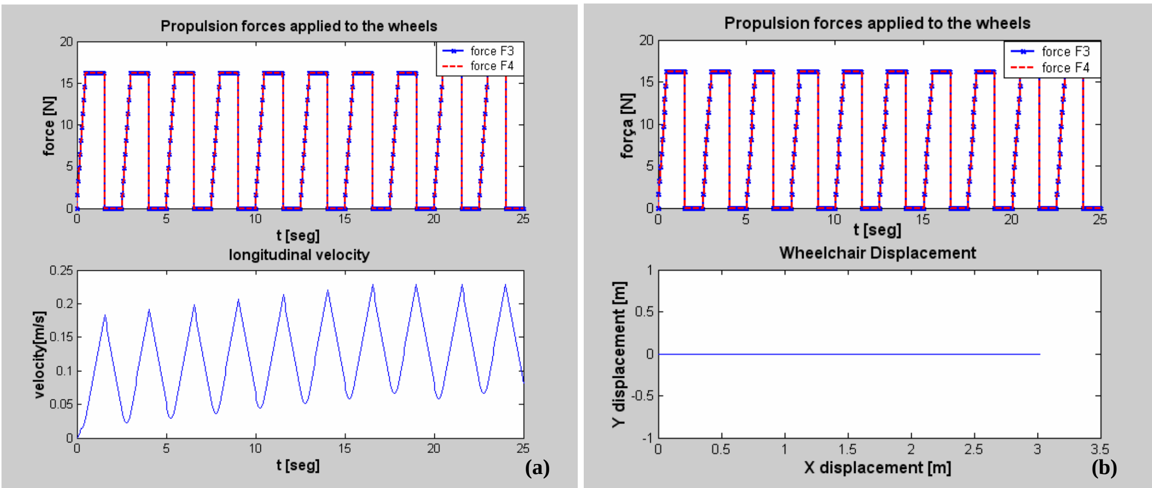


Figure 4. Wheelchair longitudinal velocity (a) and Wheelchair displacement (b)

The propulsion force provided by the user was assumed as equal and enough for the movement, (F3 = F4 = 16.18 N) for the first simulation. It can be observed, in this simulation, that the system has a linear displacement based on this assumption.

It can be observed by fig. 4 (a) that wheelchair has an irregular behavior in terms of longitudinal velocity (forward movement) caused by the propulsion force. When the propulsion force is greater than zero the wheelchair tends to increase its velocity; when this does not occur, the system trend is to decrease the velocity because of the rolling resistance force. The transverse velocity is zero, which means that the wheelchair has no lateral sliding movement.

Fig.4 (b) represents the wheelchair displacement when the propulsion forces are equal. There is no displacement deviation, because, as mentioned before, the wheelchair system only changes its moving direction when different forces are applied to the rear (or motor) wheels. The velocity decrease corresponds to the lapse of time necessary to return the arm to the initial contact point. During this time the system is under the reaction forces to movement. Summarizing the maximum displacement of the system was 3.0m in longitudinal direction and the maximum velocity was 0.23 m/s.

5. Sagittal disparity

It is a general rule, both for children and adults, that there is a small difference between the force capacity exercised by the upper members on both sides of the body. For wheelchair users this difference will result in a path deviation expected to the side of lower physical capacity.

The sagittal disparity can be associated with hemiplegia which is a more severe physical disability. Hemiplegics have more locomotion difficulties with a common manual wheelchair, because during displacement the system will need more interference for path correction, otherwise the system will accomplish an undesirable displacement.

Hemiplegia is mainly common to encephalopathy, very common in children, as cerebral palsy, which is a cerebral damage resulting from the lack of oxygen during birth and in adults resulting from a cerebral vascular accident. (Bobath et al., 1979)

Although cerebral palsy is a disorder of the muscle control system, which results from some damage to parts of the brain. The term cerebral palsy is used when the problem occurred to a developing brain either before birth, around birth or in early life. Children can have problems such as weakness, stiffness, slowness, shakiness and difficulties with balance. These problems can range from mild to severe. (Bobath et al., 1979)

A chronic encephalopathy is not progressive; it is a stationary lesion in the brain, and therefore a stable neurological status does not mean the child does not have its own evolution, acquiring abilities or overcoming stages of the neuropsychomotor evolution. (Bobath et al., 1979)

As already described the propulsion cycle will present one side where the propulsion force will be higher, in other words, the user's sides will be accomplishing the maximum effort, but the value of the force will be different.

For simulation purposes, a user will be considered with a difference of 10%, so that the left side will act with a propulsion force 10% higher than the right side, that is, the propulsion force on the left side (*F3*) will be 17,81 N while on the right side (*F4*) it will be 16,18N. The other characteristics of the propulsion cycle will remain unaffected. The propulsion forces are presented in Fig. 5

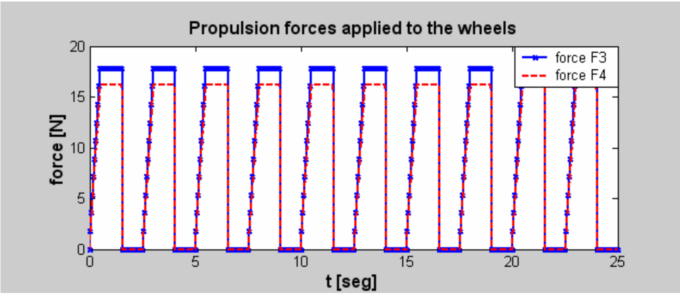


Figure 5. Propulsion Cycle with 10% of sagittal disparity

Fig. 6 (a) shows the angular acceleration and angular velocity of the system when it is under a propulsion force with a 10% sagittal disparity. In the previous simulation these two variables were zero because the system was under the same propulsion force on both wheelchair sides. What is important to show is that when the propulsion forces are equal, the momentum sum represented by eq. (7) is zero, and the angular acceleration also becomes zero. On the contrary, when angular accelerations are different from zero and because of the cyclical characteristic of the propulsion force, both acceleration and angular velocity present a cyclical behavior. When the angular acceleration is not equal to zero the system will suffer a path deviation as presented on Fig. 6(b).

Fig. 6 (b), shows that a small difference between the forces can considerably change the system displacement and its direction. For the 10 % of sagittal disparity, with the force over the left side 10% higher than over the right side, the maximum displacement obtained was 3.31 m in longitudinal direction and -0.62 m in transverse direction and the maximum longitudinal velocity was 0.23 m/s. This path deviation causes a constant path correction which will lead to faster muscular fatigue of the user, decreasing his independent mobility skill and increasing the risk of developing repetitive effort lesion.

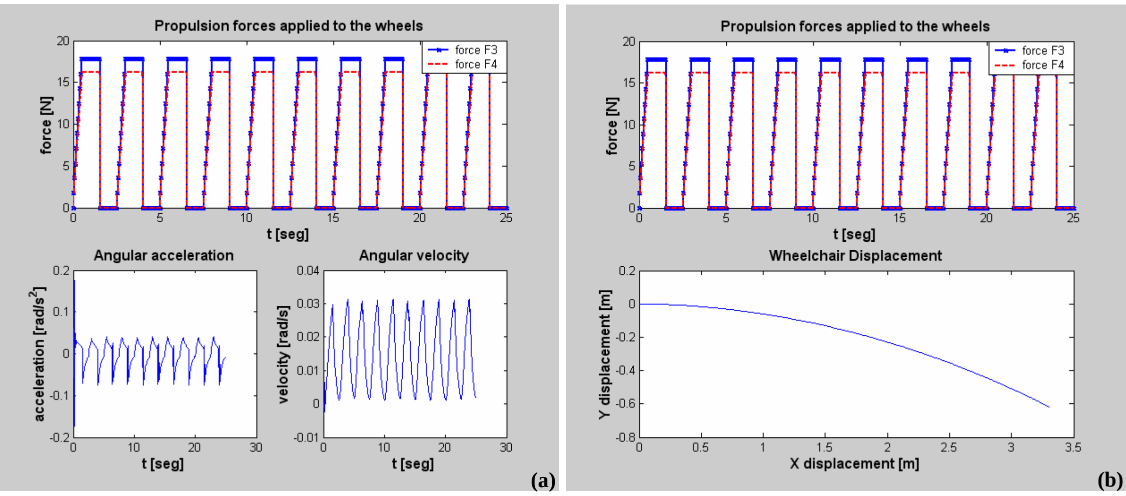


Figure 6. (a) Wheelchair angular acceleration and velocity (b) wheelchair displacement

6. Propulsion Cycle delay

The second propulsion force deformity is the cycle delay, which means that the propulsion force is applied over the propulsion ring in different instants of time.

This alteration in propulsion force can also be related to encephalopathy as with sagittal disparity or even with low muscular capacity and with the lack of user training with conventional manual wheelchairs.

This kind of deformity can be deliberate if the user wants to perform a change in the wheelchair movement direction without stopping the movement. As shown in section 4 the user needs to apply the propulsion force simultaneously and with the same intensity for the system displacement to be linear, otherwise the system will suffer deviations in the movement direction.

For simulation purposes the time delay was assumed as being equal to 0.5 s, which means that 0.5 s after the left side propulsion cycle is started, the right side propulsion cycle begins. This time delay is presented on Fig. 7, and it should be noted that the other propulsion cycle variables remain equal to the standard propulsion cycle defined previously.

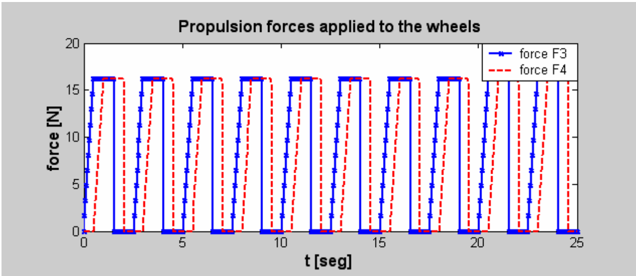
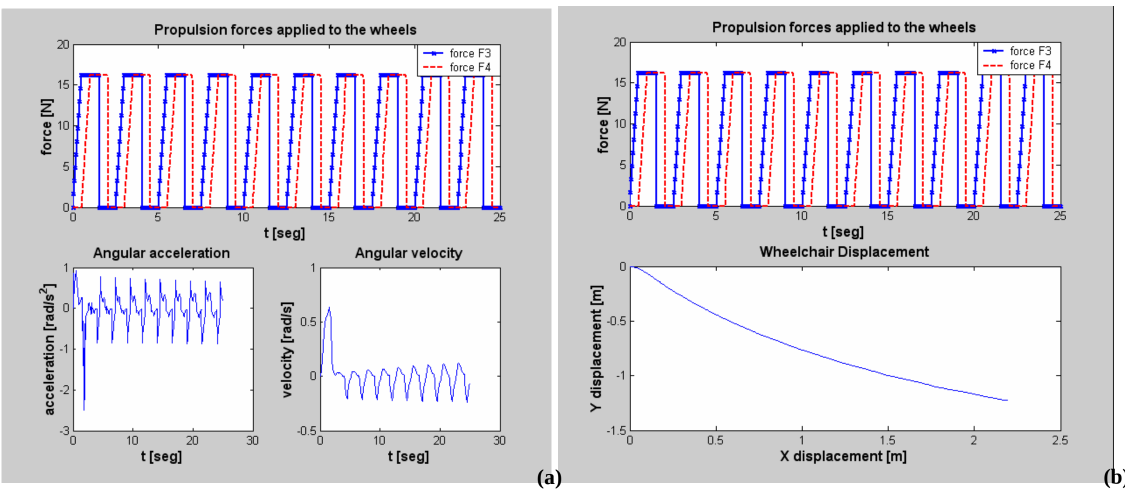


Figure 7. Propulsion Cycle with 0.5 second of time delay

As already mentioned, the system angle acceleration is zero only when both sides of the system are submitted to the same propulsion force. For sagittal delay the interval when this occurs is smaller and for this reason the system displacement deviation is higher as presented in Fig. 8(b). Due to the time delay of the propulsion cycle both acceleration and maximum velocity reach lower levels and this is coherent because if the resistance force to rolling remains constant, the result of eq. (5) becomes smaller due to the lack of synchronism of the propulsion forces.

Fig. 8(a) represents the angular acceleration and angular velocity of the system when it is under propulsion time delay of 0.5s. In the previous simulations these two variables presented lower values. It is important to show that when the propulsion forces are not equal the momentum sum represented by eq. (7) is not zero and consequently the angular acceleration also becomes different from zero. Because of the cyclical characteristic of the propulsion force, both acceleration and angular velocity also present a cyclical behavior. When the angular acceleration is not equal to zero, the system will suffer a path deviation as presented on Fig. 8(b).

One explanation for the angular acceleration and velocity values to become higher than the values found in the previous section is that the propulsion force differences caused by the time delay are higher than the difference caused by the sagittal disparity.



(a) (b)

Figure 8. (a) Wheelchair angular acceleration and velocity (b) wheelchair displacement

Fig. 8(b) shows that a small time delay between forces can also considerably change system displacement and direction. For a time delay of 0.5 s as defined before, the maximum displacement was 2.2 m in longitudinal direction and -1.23 m in transverse direction and the maximum longitudinal velocity was 0.19 m/s. This path deviation causes constant path corrections which will lead to a faster muscular fatigue of the user, decreasing his independent mobility skill and increasing the risk of developing repetitive effort lesion. Another important observation is that the displacement deviation caused by the time delay is higher than the deviation caused by sagittal disparity. This leads to the conclusion that time delay is more harmful to the system performance than sagittal disparity.

7. Propulsion Cycle with Sagital Disparity and Time delay

After becoming aware of the system behavior under the separate influence of sagittal disparity and time delay it is interesting to know what kind of propulsion cycle deformity is worse for the system behavior or, in other words, which one will prevail during displacement. In order to perform this test the system is going to be simulated again with 10% of sagittal disparity and at the same time with 0.5s time delay with both acting on the same side of the user's propulsion force, because it is a tendency for the user to have a weaker side with less motor control as well. The propulsion force resulting from the interaction of these two deformities is shown on Fig. 9.

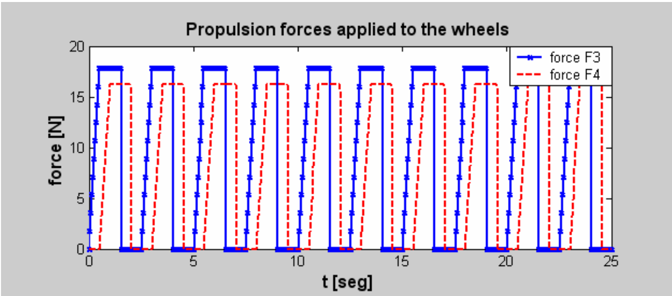


Figure 9. Propulsion Cycle with 10% sagittal disparity and 0.5 second time delay

As already mentioned, the system suffers more influence from the time delay than from sagittal disparity, as it can be observed in Fig.10(a) with both angular velocity and acceleration being very similar to the previous section results (time delay simulation). The same occurs for the other variables as longitudinal velocity and acceleration. For the displacement the behavior is similar to the time delay simulations, but the system reached a longitudinal displacement of 2.33 m and transverse displacement of 2.94 m and the maximum longitudinal velocity was 0.23 m/s. Although the system behavior is similar to the time delay, the sagittal disparity also influences the system behavior, and that is why the values are higher than in the previous section. But observing the system behavior it can be confirmed that the higher influence is from the time delay, because the system behavior is more similar to this kind of propulsion cycle deformity.

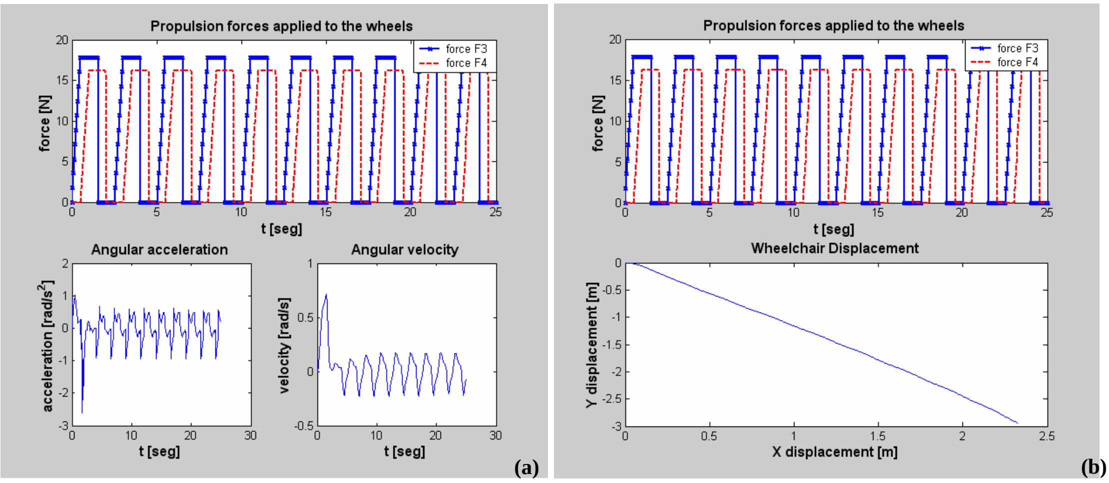


Figure 10. (a) Wheelchair angular acceleration and velocity (b) wheelchair displacement

8. Conclusion

To conclude it is important to observe that any kind of deformity presented in the propulsion cycle will result in a system deviation. Every time this happened path corrections were needed. Furthermore, accidents such as collisions with pedestrians or objects may also happen, due to the lack of movement control. Constant path corrections require a greater physical effort by the user to reach his objectives and consequently leading to a higher risk of developing DORT and reduction of his independent mobility.

The following table 1 summarizes the simulation results for the standard propulsion cycle and for the simulations under propulsion cycles with deformities

Table 1 – Summarizing simulation results

	Standard Cycle	Sagital Disparity	Temporal Delay	Both
longitudinal dispalcement	3.0297	3.3112	2.1975	2.3297
transversal displacement	0	-0.6213	-1.2278	-2.943
Max. Longitudinal velocity	0.2282	0.2304	0.1914	0.23
Max. Longitudinal Accel.	0.1472	0.1619	0.1472	0.1619
Max. Angular Accel.	0	0.1759	0.9154	1.0129
Max. Angular. Velocity	0	0.0314	0.6354	0.7166

The system suffers higher influence from the time delay as it can be observed on tab.1, with path deviation, angular acceleration and angular velocity showing higher values than sagittal disparity.

It is also important to observe that both sagittal disparity and time delay are strategies used by users to change the wheelchair direction and they do not reduce the system performance when they are applied according to the user's desire. The aim of this work is not to eliminate these deformities for every situation, but only when these deformities decrease the system performance.

The knowledge of the behavior of the system under the propulsion cycle deformities can be very useful for the analysis and specification of the type of disability of a wheelchair user, only by the observation of the system path deviation during a propulsion cycle.

Another important contribution of this work is the possibility to develop control strategies for the system that aid the user to correct the deformities of the propulsion cycle and increase the system efficiency and consequently improving the user's quality of life.

Finally, with the model employed in this work the control strategies can be tested and implemented more quickly and with a lower cost once the model proved very efficient in the reproduction of the dynamic behavior of the studied system.

9. Acknowledgements

The authors would like to thank Unicamp and Capes for the financial support.

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8. Responsibility notice