INFLUENCES OF THE FORCE OF PROPULSION WHEELCHAIR STANDARD OVER THE WHEELCHAIR VELOCITY

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Abstract. *The aim of this article is to relate propulsion force characteristics with wheelchair velocity. Thus therapists can use it to develop physical training to increase user's pulmonary or muscular capacity. The dynamic model for the system was constructed based on a double bicycle model with driving force applied over the rear wheels and steering on the front wheels, which is a coherent model to determine the system behavior as mentioned in the literature. This model has as input variables wheelchair geometric and constructive characteristics such as width, center of gravity localization, wheel rigidity, friction coefficient. Furthermore, the model considers the propulsion force applied by the user as the system driving force. The propulsion force used to simulate the model was an arc path over the propulsion ring and a trapezoidal behavior which can be divided in three sections: one constant increasing section, one constant value section and another null section. The system was simulated varying the values for the time spent for these three sections and as result it was observed that the system velocity increases with the decrease of the time spent by the first and second sections but there is a maximum value for the second section which corresponds to the lapse of time equal to one second.*

Keywords*: Biomechanics, Propulsion Force, Dynamic Behavior*

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.

For this work it was considered that over the wheelchair propulsion ring the upper members of the user describe a path equal to a circumference arch, the behavior of the propulsion force module has a trapezoidal form and can be divided in three sections, the first where the propulsion force grows continuously from null to the maximum value, the second where the maximum value is maintained for a certain period of time and the third where the force becomes null.

Of course the velocity behavior is related with system characteristics such as weight and rolling resistance. If they are maintained constant however the velocity behavior can be modified through the characteristics of the propulsion force, or in other words, of the time spent in each of the sections comprising this force.

The maximum propulsion force applied over the propulsion ring can also change the velocity behavior, but this characteristic is only dependent on the user muscular capability. For a velocity increase it is more important to decrease the period of time spent in the propulsion cycle than increasing the force applied.

Therapists can use the relation between the propulsion force characteristics and wheelchair velocity to develop physical training to increase users' pulmonary or muscular capacity. This knowledge is very important to develop a training program for maximum propulsion force efficiency, as is the case for the training of para-olympic athletes.

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 maximum propulsion force and time characteristics of the propulsion cycle.

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.

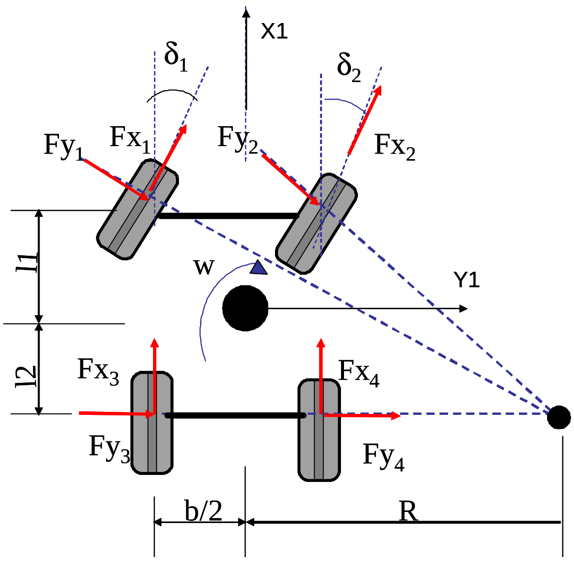

Figure 1. Free Body Diagram for the wheelchair

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 $\dot{Vx}$, $\dot{Vy}$ e $\dot{w}_z$, respectively longitudinal, transverse and angular accelerations of the system. The equations resulting from this procedure are the following:

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

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

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

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*, due to the lack of wheelchair reference material.

3. Considerations about the propulsion force

The propulsion force is the form of interaction of the user with the wheelchair and also with the control system. For this reason, its definition and the knowledge of its behavior is extremely important for the success of any control strategy.

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*.

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).

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

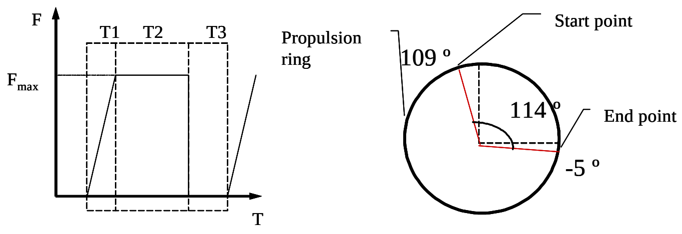

Figure 2. Characteristics of Propulsion Force definition

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.

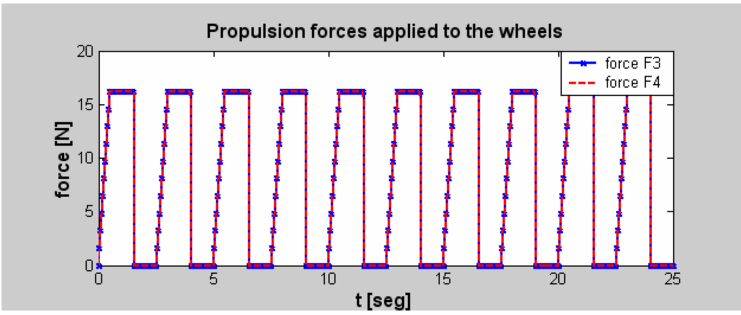

Figure 3. Standard propulsion cycle

It is important to note that in the simulations the propulsion cycle will not suffer any deformation such as sagittal disparity or time delay, only variables *T1*, *T2* or *T3* will be changed.

4. Manual wheelchair simulation – Standard situation

The aim of this section is to study the dynamical behavior of a wheelchair under the influence of a standard propulsion cycle. This information is important because based on the results it will be possible to quantify the efficiency improvement when using a servo-assisted motorization system or even quantify the changes when the system is under the influence of a propulsion cycle away from the pattern.

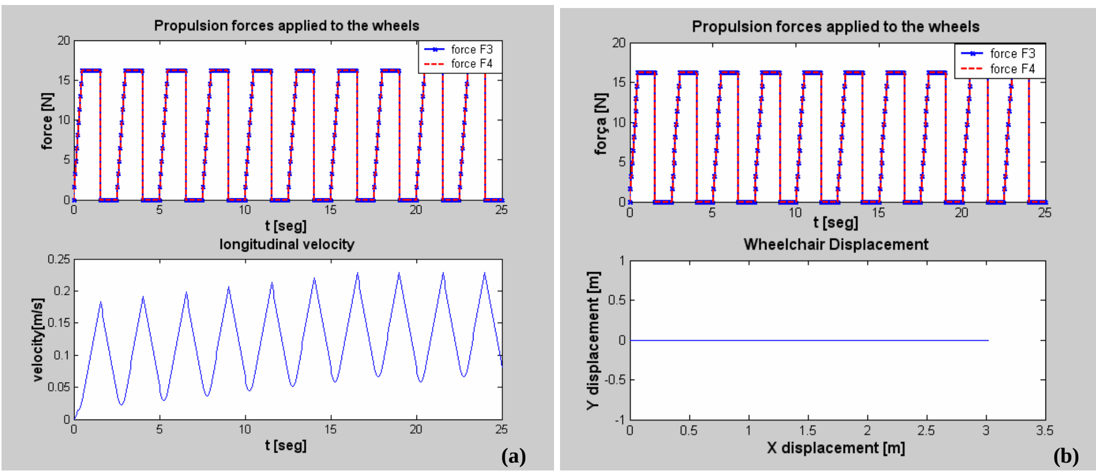

Figure 4. Wheelchair longitudinal velocity (a) and Wheelchair displacement (b)

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. Influences of variables *T1, T2* and *T3* over wheelchair velocity

Using the model developed in section 2, the system will be simulated for several values of T1, T2 an T3, resulting in 3D graphics that will show the influence of each variable over the wheelchair maximum velocity.

The first to be studied is variable *T1*. This variable represents the lapse time necessary for the user to apply the force beginning at zero until the maximum value (*Fmax*), as already mentioned. Variable *T1*, does not depend on the user wishes, because it is associated to physical factors as lesion level, age, sex, pulmonary capacity as established by Dallmeijer et al. (1994) and Kulig et al. (2001).

This variable is associated to user capability to apply power during the propulsion cycle and for this reason through physical and training exercises a therapist can reduce this time increasing the efficiency of the propulsion cycle. The following figure 5 represents the wheelchair velocity under the influence of variable *T1*. Variable *T2* was considered equal to 1s and variable *T3* equal to 1s. These values correspond to the standard values defined previously.

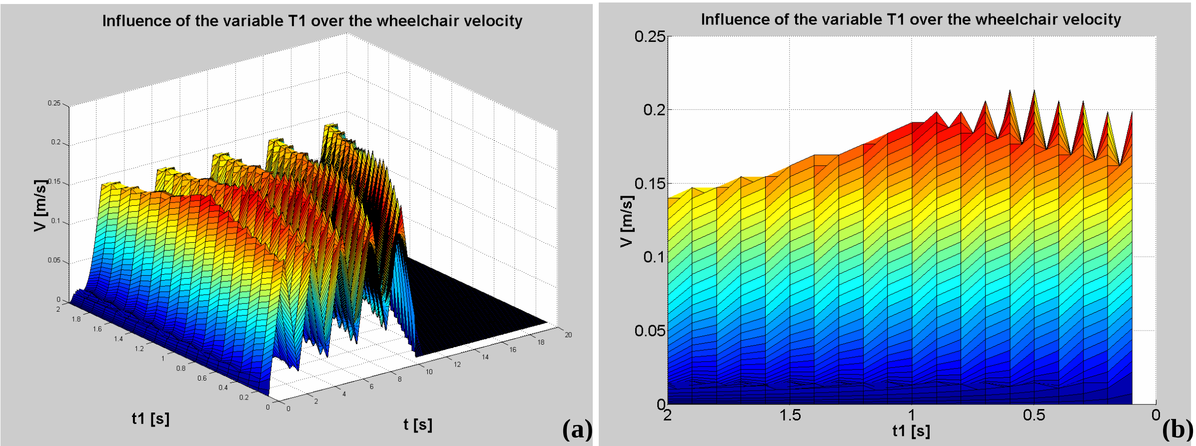

Figure 5. (a) 3D graphic of the influence of variable *T1* over the wheelchair velocity, (b) plan view of the influence of variable *T1* over the wheelchair velocity

Varying the values of any variable (*T1*, *T2* or *T3)* the lapse of time for the complete propulsion cycle also changes, because the total lapse of time for a complete propulsion cycle is defined as being equal to (*T1+T2+T3*). It can therefore be observed that in fig. 5(a) there is a time period in which the velocity becomes zero, because repetitions of the propulsion cycle used in the simulations end with a smaller time period than *T1*. Observing fig.5 (b) it can be concluded that the velocity presents a maximum value when T1 is approximately equal to 0.6 s.

This maximum point, around 0.5 and 0.6 seconds, has an important meaning because coaches or therapists can train their athletes to drive their wheelchairs with T1 next to this value in order achieve maximum efficiency.

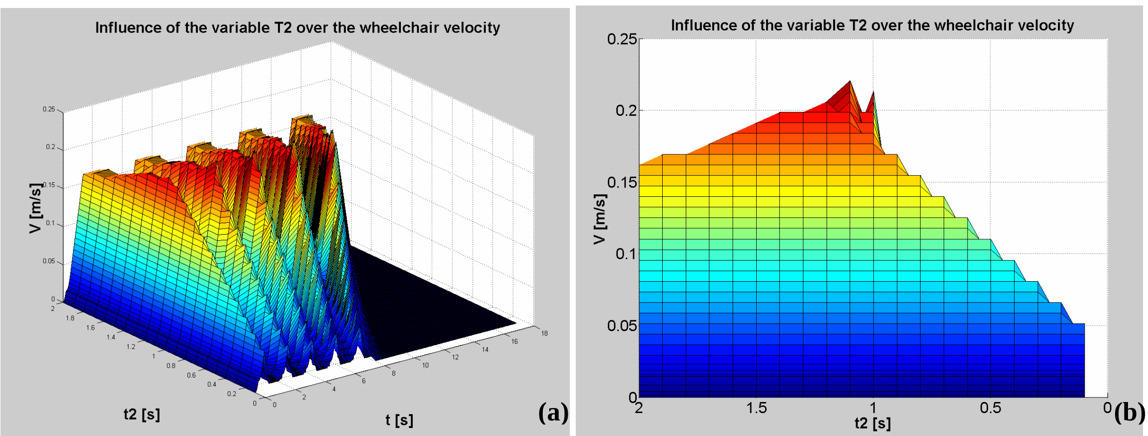

Figure 6. (a) 3D graphic of the influence of variable *T2* over wheelchair velocity, (b) plan view of the influence of variable *T2* over wheelchair velocity

Variable *T2*, represents the time when the propulsion force remains constant up to the moment the user releases the hand from the propulsion ring to return to the initial contact point. This variable is dependent on the user's physical capacity and also the total extension of the propulsion cycle. The total extension of the propulsion cycle corresponds to the distance measured in degrees over which the user is able to apply force on the propulsion ring , as shown on fig. 2.

The results presented in fig. 6 (a) were obtained adopting *T1* equal to 0.5 s, *T3* equal to 1 s and *T2* assumed values between 0.1 and 2 seconds. Again, observing fig. 6(b), the maximum velocity occurs at a specific value, *T2* equal to 1.1 seconds, and this value represents the point where the velocity is higher and consequently the maximum propulsion cycle efficiency.

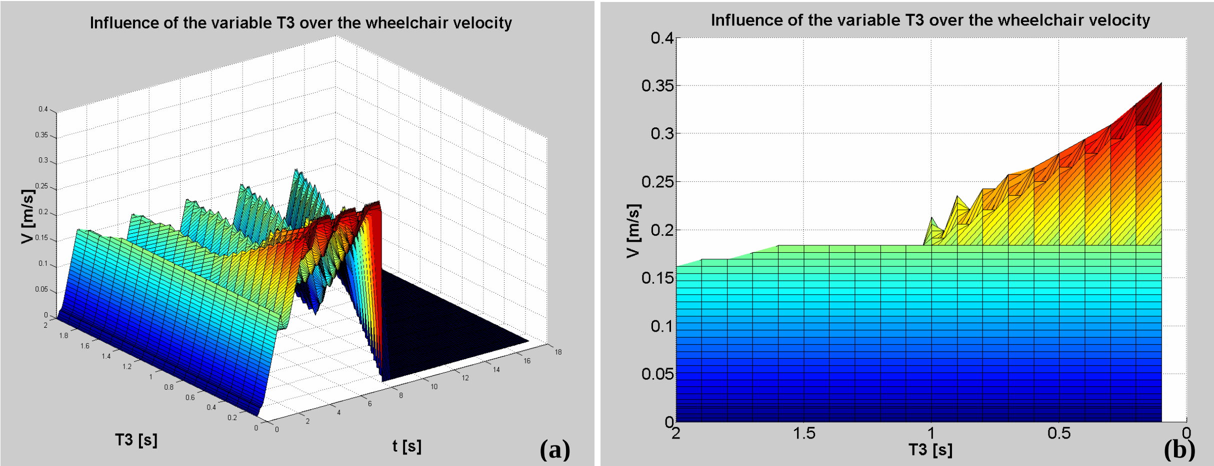

Figure 7. (a) 3D graphic of the influence of variable *T3* over the wheelchair velocity, (b) plan view of the influence of variable *T3* over the wheelchair velocity

Variable T3, represents the time necessary for the user to return the hand to the initial contact point on the propulsion ring to repeat the driving cycle. This time depends only of the physical capacity of the user, because it represents the time necessary for muscle relaxing to retake the propulsion force application.

For the simulation presented in fig. 7 (a) adopted *T1* equal to 0.5 s, *T2* equal to 1 s and *T3* assumed values between 0.1 and 2 seconds. Contrary to the observed in fig. 7(b), the maximum value for the wheelchair velocity occurs when *T3* is equal to 0.1s, but analyzing the result it can be concluded that the maximum propulsion cycle efficiency occurs for the minimum value of T3. Propulsion cycle efficiency is, therefore, highly dependent on user physical performance.

6. Influences of variable *Fmax* over wheelchair velocity

In order to confirm the assumption that variable *Fmax*, which represents the maximum propulsion force applied by the user, has no significant influence over the system maximum velocity, will be simulated adopting values for *Fmax* from 8,0N to 23,5 N. The result of this simulation is presented in fig.8.

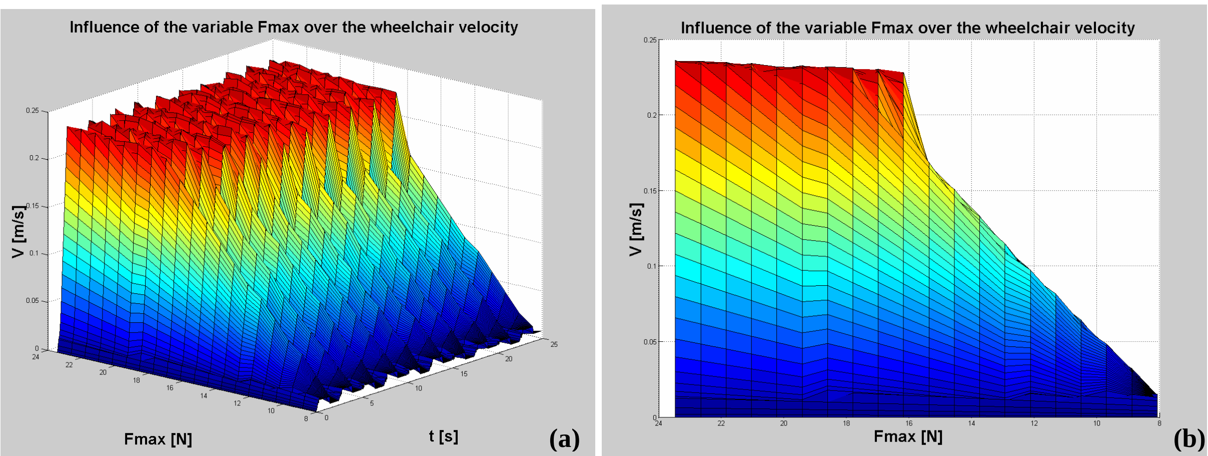

Figure 7. (a) 3D graphic of the influence of variable *T3* over the wheelchair velocity, (b) plan view of the influence of variable *T3* over the wheelchair velocity

It can be observed that after the user applies a minimum force for the system start movement, *Fmax* equal to 16.18N, the maximum velocity reached by the system did not suffer any great increase although the propulsion force reaches 23,5 N which represents an increase of 50%¨related to the minimum propulsion force. Obviously, if *Fmax* is lower than the minimum value, the system velocity will increase proportionally to the value of *Fmax*.

7. Conclusion

It can be concluded with this work that it is not only increasing wheelchair user physical capacity that will0 increase wheelchair velocity and consequently the propulsion cycle efficiency. Physical training is important to decrease variable *T3*, which represents the user physical capacity and with its reduction the higher will be the velocities reached by the user. On the other hand, for variables *T1* and *T2*, which have a specific value that corresponds to a maximum velocity, only the increase in physical capacity is not enough to reach maximum efficiency. A training is therefore necessary to adjust the user propulsion cycle to get near the determined values of 0.6 s and 1.0 s respectively.

The most efficient propulsion cycle for the simulations in this work presents as characteristic the following values: *T1* equal to 0.5 s, *T2* equal to 1.1 s and *T3* equal to 0.1s. Excepting variable T3, all the others are very close to the standard propulsion cycle defined in the literature. It is also important to note that for T3 equal to 0.1 s the user will be able to achieve this performance only during very short period of time and for longer distances higher values than these will be adopted for this variable.

It is important to note that although the values obtained in this work are close to those found in the literature and through the observation of wheelchair users, in most cases they are not aware that their propulsion cycle is more or less efficient. The wheelchair user adapts to a propulsion cycle that is more in accordance with their needs and abilities, and as shown in this work in most cases these characteristics are close to the maximum efficiency point.

Finally with the model presented in this work and the results reached paraolympic coaches can improve the performance of their athletes based on the dynamic model and most importantly without the overload on the upper member muscles and with the approximation of the propulsion standard of the athlete to the point of maximum efficiency. This change of training focus can reduce the risk of damages for repetitive efforts (DORT), which will jeopardize the athlete's performance.

8. Acknowledgements

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

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8. Responsibility notice