Velocity Control for Walk Assistance by Endeffector Force in the Leg Coordinate based on the Biarticularly-actuated System

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Abstract— **This paper proposes a novel velocity control of the center of mass (COM) of a human body with attached ankle foot orthosis (AFO) during the stance phase. We propose a novel coordinate system for COM that achieves model simplificatio with the biarticularly-actuated system. This allows for simple control design of the velocity and position of COM. In addition to simplifie control, the proposed mathematical model for AFO has a simple structure that reproduces the biarticularly-actuated system using passive elements such as springs. Simulation results and comparison with conventional methods verify the effectiveness of the proposed control design.**

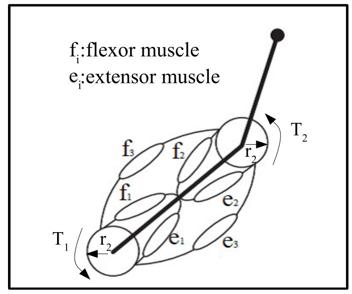

Fig. 1. 3 pairs 6 muscles model

I. Introduction

Hemiplegia is known as a sequela related to cerebrovascular diseases, such as apoplexy. People who are suffering from hemiplegia fin walking difficult Especially in case of elderly people hemiplegia imposes a danger of fatal injuries which include joint dislocation and fracture, for them being easy to fall to the paralyzed side during walking.

In this paper, we perform coordinate conversion using biarticular actuation, and propose a simple velocity control method based on the end-effector force control for a two-link manipulator. Based on this result, we are fabricating the new AFO which can ease the patients' burden by substituting the patients' gastrocnemius muscle with a passive element.

Various kinds of AFO have been developed for hemiparetic gait assistance. Blaya et al. developed AFO with variable impedance of an ankle, which prevents foot drop during stance phase, aggravation of symptoms, and accidents [1]. Yamamoto et al. developed a hydraulic AFO with variable rigidity of the ankle, which assists dorsifl xion and plantarfl xion during walking [2]. However these orthoses only help the patients who can walk by themselves. Thus, there are needs for AFOs which generate propulsion force for forward walking for more severely disabled people.

Some peculiar muscles called biarticular muscles which characterize animal limbs are receiving attention in recent years. Biarticular muscles are attached ranging over two joints, and drive the two joints simultaneously. Conventional robotics, however, seldom takes biarticular muscles into consideration despite the fact that they are the essential actuators in animal limbs. Kumamoto and Hogan verifie the effectiveness of biarticular muscles in end-effector force characteristics using the animal limb model [4],[5] using biarticular actuators.

Animals with limbs perform various movements such as walking, running, and jumping by harnessing relevant muscles. These animal movements can be described by a spring model called SLIP (Spring Loaded Inverted Pendulum) model [3]. The SLIP model implies that it is not one joint or one muscle, but mutually working multiple joints and muscles that achieves animal locomotion. This property inspired many research works on biarticular muscles. For example, Lewis et al. tried to reproduce the dynamic movement of animals driving two joints of the legs by using only one motor [6], Iida et al. used a spring between the joints of the leg [7]. And Klein et al. tried to reproduce the biarticular muscle movement during human walking using belt [8]. As seen in the aforementioned research works, the consistency in animal movements are explained by applying biarticular muscles.

In Section Ⅱ, biarticularly actuated drive mechanism is shown. In Section Ⅲ, we defin a supporting leg model during walking, and a two-link manipulator in consideration. Then we defin the leg coordinate system which is fi ed to Link 1. Based on this coordinate system the kinematics and statics are described. In Section Ⅳ, we propose a velocity control method of the center of mass using the end-effector force control in the leg coordinate system. In Section Ⅴ, a simplifie method is examined for the limb model using passive biarticular elements. Finally Section VI concludes the work.

II. Biarticularly-actuated model and Mathematical expression of joint torques

Animal bodies are driven by various muscles. However if we restrict limb movements into planar ones, they can

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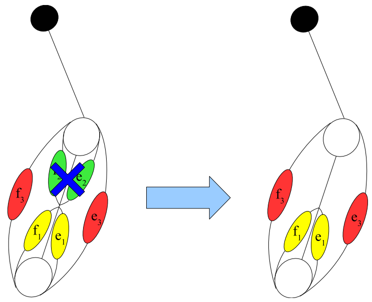


Fig. 2. Mono-Bi system without τ_2

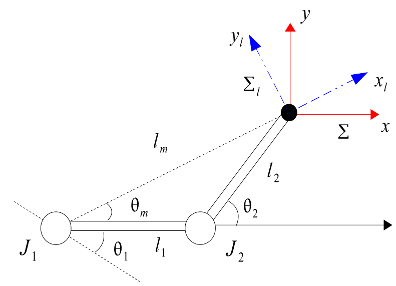


Fig. 3. Definitio of novel leg coordinate

be expressed with the 3 pairs 6 muscles models as shown in Fig.1 where $f_i, \{i = 1, 2, 3\}$ represents extensor muscles and $e_i, \{i = 1, 2, 3\}$ represents fl xor muscles. f_1, e_1 are monoarticular muscles of the firs joint, f_2, e_2 are monoarticular muscles of the second joint, and f_3, e_3 are biarticular muscles over two joints. These are driven as a pair, and if torques that each pair exerts are define as $\tau_i\{i = 1, 2, 3\}$, then the joint torques T_1, T_2 are denoted as the following formula (1) (2).

$$T_1 = \tau_1 + \tau_3 \quad (1)$$

$$T_2 = \tau_2 + \tau_3 \quad (2)$$

Since there are three inputs and two outputs as shown in the formulae above, biarticular-muscle-driven system has redundancy. However, Oh et al. clarifie that this redundancy contributes greatly to direction control of end-effector force [9]. It was shown that with biarticular muscles it is easy to realize straight forward movements since the straight line connecting the firs joint and end-effector ease the control.

III. Coordinate System and Kinematics Using Biarticular Muscles

In human walking behavior, the velocity of COM is known to be obtained by the propulsive force exerted by the muscles of the leg [4]. In this paper, we focus on generating propulsive force which pushes the COM using the supporting leg with an AFO during stance phase of a patient with hemiplegia. In definin the supporting leg model during stance phase, we propose the Mono-Bi system shown in Fig.2 which is equipped only with a monoarticular muscle pair τ_1 of the firs joint and a biarticular muscle pair τ_{12} extracted from 3 pairs 6 muscles model. That is, the torque of the monoarticular muscle pair of the second joint is zero, $\tau_2 = 0$. Moreover, with such actuator configuration we also propose the leg coordinate system which is suitable for describing movements at the end-effector of a manipulator.

A. Proposal of the Leg coordinate suitable for biarticular muscle inputs

J_1 and J_2 represent the firs and the second joint, and l_1, l_2 are the lengths of each link. Joint angles are denoted as θ_1 and θ_2. m_1, m_2, and I_1, I_2 are the mass and the inertia of each link member. Joint torque is denoted as T_1 and T_2. And finall the mass of the point P, the COM of the upper body is written as m_0.

B. Derivation of kinematics and statics in the proposal Leg coordinate

In the case of two link manipulator, it is common to use the absolute coordinate system. In this paper, however, we propose a coordinate system which is fi ed to the tip of the end-effector. When considering two muscles present in this work, it becomes easier to discuss the movements of end-effector P using this kind of coordinate system. We call this Σ_l coordinate leg coordinate system.

Link lengths of the manipulator shown in Fig. III-B are as mentioned above, the length of the straight line which connects firs joint J_1 and end-effector P is l_m, θ_1 is an angle between the absolute coordinate and the Σ coordinate system, where $\theta_{12} = \theta_1 + \theta_2$. $\Delta x_l, \Delta y_l$ are the displacements of each axis, and $\Delta\theta_1, \Delta\theta_{12}$ are the displacement of angles θ_1 and θ_{12}. For simplification each length is set in a way that $l_1 = l_2 = l$. The kinematics and inverse kinematics in the leg coordinate are written as follows (3) (4).

$$\begin{aligned}\begin{pmatrix}\Delta x_l\\ \Delta y_l\end{pmatrix} &= \frac{1}{l_m}\begin{pmatrix} l^2\sin\theta_2 & -l^2\sin\theta_2\\ l^2(1+\cos\theta_2) & l^2(1+\cos\theta_2)\end{pmatrix}\begin{pmatrix}\Delta\theta_1\\ \Delta\theta_{12}\end{pmatrix}\\ &= \frac{l^2}{l_m}\begin{pmatrix}\sin\theta_2(\Delta\theta_1-\Delta\theta_{12})\\ (1+\cos\theta_2)(\Delta\theta_1+\Delta\theta_{12})\end{pmatrix} \quad (3)\end{aligned}$$

$$\begin{aligned}\begin{pmatrix}\Delta\theta_1\\ \Delta\theta_{12}\end{pmatrix} &= \frac{1}{l_m\sin\theta_2}\begin{pmatrix}\cos\theta_2+1 & \sin\theta_2\\ -\cos\theta_2-1 & \sin\theta_2\end{pmatrix}\begin{pmatrix}\Delta x\\ \Delta y\end{pmatrix}\\ &= \frac{\cos\theta_2+1}{l_m\sin\theta_2}\begin{pmatrix}1\\ -1\end{pmatrix}\Delta x + \frac{1}{l_m}\begin{pmatrix}1\\ 1\end{pmatrix}\Delta y \quad (4)\end{aligned}$$

Torque exerted by the monoarticular muscle of the firs joint is τ_1, and one exerted by the biarticular muscle is denoted as τ_3. The force outputs of the P in the leg coordinate are f_l^x and f_l^y.

$$\begin{aligned}\begin{pmatrix}\tau_1\\ \tau_{12}\end{pmatrix} &= \frac{1}{l_m}\begin{pmatrix} l^2\sin\theta_2 & l^2(1+\cos\theta_2)\\ -l^2\sin\theta_2 & l^2(1+\cos\theta_2)\end{pmatrix}\begin{pmatrix}f_l^x\\ f_l^y\end{pmatrix}\\ &= \frac{l^2\sin\theta_2}{l_m}\begin{pmatrix}1\\ -1\end{pmatrix}f_l^x + \frac{l^2(\cos\theta_2+1)}{l_m}\begin{pmatrix}1\\ 1\end{pmatrix}f_l^y \quad (5)\end{aligned}$$

$$\begin{aligned}\begin{pmatrix}f_l^x\\ f_l^y\end{pmatrix} &= \frac{1}{l_m\sin\theta_2}\begin{pmatrix}\cos\theta_2+1 & -\cos\theta_2-1\\ \sin\theta_2 & \sin\theta_2\end{pmatrix}\begin{pmatrix}\tau_1\\ \tau_{12}\end{pmatrix}\\ &= \frac{1}{l_m\sin\theta_2}\begin{pmatrix}(1+\cos\theta_2)(\tau_1-\tau_{12})\\ \sin\theta_2(\tau_1+\tau_{12})\end{pmatrix} \quad (6)\end{aligned}$$

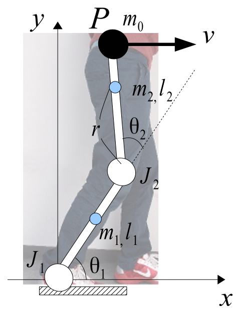


Fig. 4. Lower extremity modeled as 2 link manipulator

As mentioned above, f_l^x, f_l^y, the outputs in the leg coordinate system are obtained by the difference and the sum mode of each joint torque, respectively.

IV. Proposal of the velocity control method using outputs of biarticular muscles

The two link supporting leg model including the upper body used in this work is shown in Fig. 4. For simplification movements of the supporting leg model are limited to a two-dimensional plane. Inclination of the upper body rarely changes when non-disabled people walk, and even in case of patients with hemiplegia, it is possible to control so that the upper body does not fall according to muscles of the trunk. This kind of control, however, is not a part of the control for the supporting leg. Thus in this paper, we consider the mass of the upper body equivalent to a point mass at P. v_{lx} and v_{ly} are the linear velocities of the COM in terms of the leg coordinate system. J_1 is regarded as ankle, and we set each link length $l_1 = l_2 = l$ for simplification J_1 is fi ed at this time since we are considering supporting phase, and movements of the COM shows a circumferential movement with a radius of l_m.

A. Velocity control using biarticulary actuation in the leg coordinate

We propose velocity control of the COM, making a supporting leg follow reference value l_m^{ref} by using outputs seen at the point P in the leg coordinate.

Using the relation of kinematics and leg length $l_m = 2l\cos\theta_m$ the actual velocity in each direction, v_{lx} and v_{ly} in the leg coordinate can be expressed as follows.

$$v_{lx} = l\sin\theta_m(\dot{\theta}_1 - \dot{\theta}_{12}) \quad (7)$$

$$v_{ly} = l\cos\theta_m(\dot{\theta}_1 + \dot{\theta}_{12}) \quad (8)$$

If we put the gain for the leg length error as K_p, and the gain for the speed error in each direction asK_{dx} and K_{dy}, respectively, then the control inputs, f_l^x and f_l^y for each direction seen at point P can be written as follows.

$$f_l^x = K_p\left(l_m^{ref} - l_m\right) + K_{dx}\left(v_{lx}^{ref} - v_{lx}\right) \quad (9)$$

$$f_l^y = K_{dy}\left(v_{ly}^{ref} - v_{ly}\right) \quad (10)$$

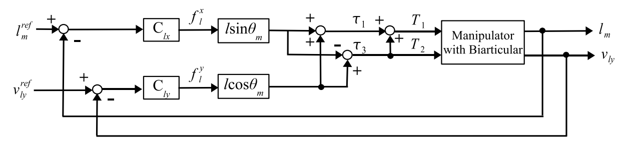


Fig. 5. Control design using leg coordinate

Note that we set $v_{lx}^{ref} = 0$ in this paper. In this way, the control input f_l^x becomes equivalent to a PD controller for the leg length l_m.

B. When biarticular muscles are not passive elements

From (9) (10), and considering $v_{lx}^{ref} = 0$, the torques of each pair of muscles τ_1, τ_{12} can be written as follows, by using the relationship (5) of the statics and leg length l_m in the leg coordinate system, then these become the input torques.

$$\tau_1 = K_p l\sin\theta_m\left(l_m^{ref} - l_m\right) + K_d l\cos\theta_m v_{ly}^{ref} - K_d l^2\left(\dot{\theta}_1 + \cos\theta_2\dot{\theta}_{12}\right) \quad (11)$$

$$\tau_{12} = -K_p l\sin\theta_m\left(l_m^{ref} - l_m\right) + K_d l\cos\theta_m v_{ly}^{ref} - K_d l^2\left(\cos\theta_2\dot{\theta}_1 + \dot{\theta}_{12}\right) \quad (12)$$

As the counterpart for comparison, the velocity reference of COM vref and the torque references (T_1, T_2) are given. As shown also in the block diagram shown in Fig. 5, because complex calculation of Jacobian matrix is unnecessary in the proposed method, a simple controller can be realized.

C. When biarticular muscles are passive elements

In addition to the input torques in (11) and (12), we assume there is a passive element which is a biarticular torque τ_{12} generating elastic torque. That is, (12) can be rewritten as follows.

$$\tau_{12} = K_{fix}\Delta\theta_{12} \quad (13)$$

Where, $\Delta\theta_{12}$ is the angle displacement from the equilibrium position θ_{12}^0, and it is define as follows.

$$\Delta\theta_{12} = \theta_{12}^0 - \theta_{12} \quad (14)$$

Moreover, gain K_d for the speed error becomes (15) below, by introducing the spring coefficien K_{fix} and considering (12) and (13).

$$K_d = \frac{K_{fix}\Delta\theta_{12} + K_p l\sin\theta_m\left(l_m^{ref} - l_m\right)}{l\cos\theta_m v_{ly}^{ref} - l^2\left(\cos\theta_m\dot{\theta}_1 - \dot{\theta}_{12}\right)} \quad (15)$$

From these observations, when Biarticular torque τ_{12} is included as a passive mechanism to movements, it becomes the input torques in (11) and (13).

TABLE I

PARAMETERS OF SIMULATION MODEL

g = 9.8[m/s^2]	m_0 = 50[kg]	m_1= 1.8[kg]
m_2= 3.6[kg]	l_1 = 0.4[m]	l_2 = 0.4[m]
I_1 = 0.217[kgm^2]	I_2 = 0.434[kgm^2]	l_m^{ref} = 0.78[m]

D. Forward walking simulation by the proposal method

Since we consider an AFO which is attached under the knee, it is assumed for the simulations that the firs joint J_1 of two link manipulator described in Section Ⅲ corresponds to the ankle of the supporting leg model. The parameters used in the simulations are indicated in table I. The initial posture is given at (θ_1, θ_2) = (1.8 [rad], 0.4 [rad]). The simulation is done from treading in until kicking out in the supporting leg model. The velocity reference of each axial direction is given in the forward direction at (v_l^{ref}, v_l^{ref}) = (0 [m/s], 1.0 [m/s]).

1) When biarticular muscles are not passive elements: In case input torques don't contain passive elements, the governing equations are written as (11) and (12). K_p, K_d can be considered as the P and D gain for the position references in the direction of x_l, respectively. These two gains are computed by pole assignment. In this case, the plant can be written as (16) below.

$$P(s) \quad = \quad \frac{1}{Ms^2} \tag{16}$$

Here, M is equal to the COM point mass m_0. Using the plant and the PD controller above, pole assignment is done. Simulation is done with a given pole at $\omega = 50$[rad/s].

2) When biarticular muscles are passive elements: In case input torques contain passive elements, the governing equations are written as (11) and (13). We use the same pole at $\omega = 50$[rad/s], and the spring coefficien at $K_{fix} = 550$[Nm/rad]. The variable gain K_d is determined by substituting gains K_{fix}, K_p into (15).

E. Simulation result

The simulation results of both cases are shown. These results are fairly consistent to that of human walking, even when the biarticular muscle τ_{12} is a passive element.

When not included a passive element, values follows well in reference values. On the other hand, we can see the characteristics of a passive element when included.

V. APPLICATION OF THE SUPPORTING LEG CHANGE ALGORITHM

In the previous section, the velocity control only for one step of the supporting leg model was considered. However, since changing legs is essential in order to extend the algorithm to walking, leg-change should be discussed. Thus here, the inverted pendulum model is introduced [10].

A. The changing algorithm of the leg in walk operation

Fig. 12 is a walking model including upper body and the leg using the inverted pendulum model. For simplification the model is limited to the sagittal plane and the mass of legs are ignored. $m, J, g,$ and l are the mass, the inertia of the upper body, the gravitational acceleration, and the length from the waist to the center of mass of the upper body, respectively. r is the length from a tip of a foot to the waist, and ϕ_1, ϕ_2 are the inclination angles of a leg and the upper body from the vertical. $r, \phi_1,$ and ϕ_2 are state variables changing with time. τ is the torque acting on the waist, and F is the generalized force corresponding to the length of the leg r.

The state equation of the inverted pendulum model is decoupled and linearized, and the poles where the system is asymptotically stable are assigned. The body can be stabilized by the pole assignment, and when ϕ_2 approaches 0, ϕ_1 has an unstable pole.

If we put the required time to take one step as T, the state variables $(\phi_1, \dot{\phi}_1)$, the inclining angle of the leg can be written as follows.

$$\begin{pmatrix} \dot{\phi}_1 \\ \ddot{\phi}_1 \end{pmatrix} = \begin{pmatrix} 0 & 1 \\ \frac{g}{r+l} & 0 \end{pmatrix} \begin{pmatrix} \phi_1 \\ \dot{\phi}_1 \end{pmatrix} \tag{17}$$

Equation (17) can be discretized by walking cycle T,

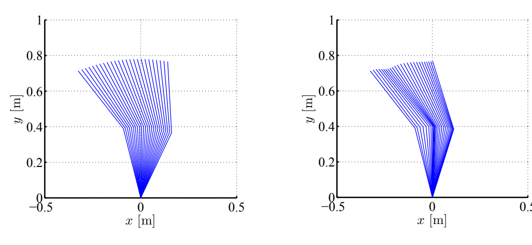
$$\phi_D[k+1] = \begin{pmatrix} \cosh bT & \frac{1}{b}\sinh bT \\ b\sinh bT & \cosh bT \end{pmatrix} \phi_D[k] - \begin{pmatrix} \cosh bT \\ b\sinh bT \end{pmatrix} u_D[k] \tag{18}$$

where, $b^2 = \frac{g}{r+l}$. This discrete system can be stabilized by determining h_1 and h_2 by assigning poles when the input is u_D from (19). Input u_D is the angle between both the legs for one step required at the instant of landing. In practice, u_D is define as (19) below.

$$u_D = h_1(\phi_{end}[k] - \frac{\phi_r}{2}) + h_2(\dot{\phi}_{end} - \frac{v - v^{ref}}{r}) + \phi_r \tag{19}$$

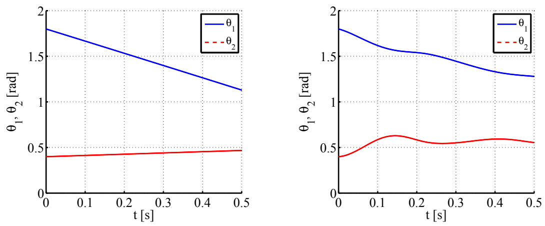
h_1 and h_2 are coefficient which assign the arbitrary poles in a discrete system, and can be written as below.

$$\begin{cases} h_1 = 1 - \lambda_1\lambda_2 \\ h_2 = \frac{(1+\lambda_1\lambda_2)\cosh bT - \lambda_1 - \lambda_2}{b\sinh bT} \end{cases} \tag{20}$$



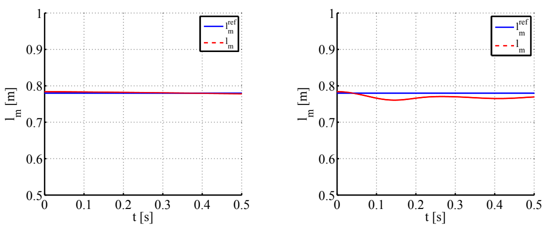
(a) W/O passive element (b) W passive element

Fig. 6. Stick diagram of supporting leg



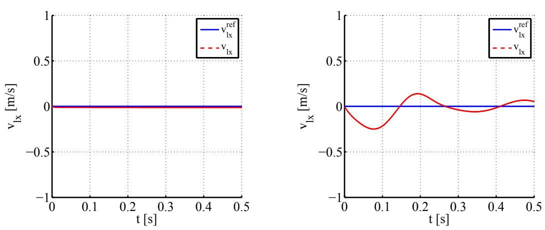
(a) W/O passive element (b) W passive element

Fig. 7. Joint angles θ_1, θ_{12}



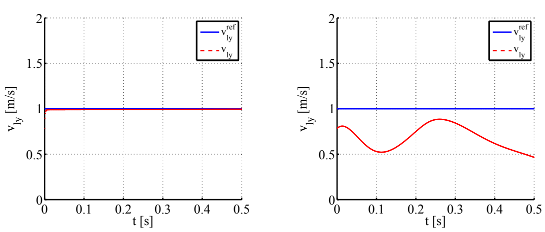
(a) W/O passive element (b) W passive element

Fig. 8. Change of leg length l_m



(a) W/O passive element (b) W passive element

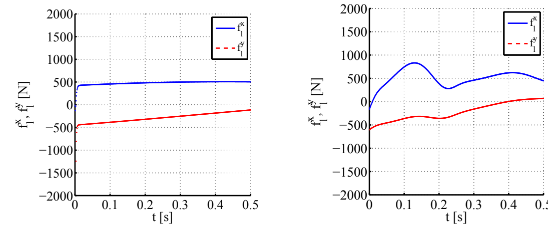
Fig. 9. Velocity of center of mass v_{lx}



(a) W/O passive element (b) W passive element

Fig. 10. Velocity of center of mass v_{ly}

ϕ_r and v^{ref} are reference step width and the velocity reference, respectively. Fig. 13 shows the leg-change model of the supporting leg.



(a) W/O passive element (b) W passive element

Fig. 11. Input forces f_l^x, f_l^y in the leg coordinate

B. State variables in changing legs

If we set ϕ_1 and r of the inverted pendulum model, as $\phi_{1,end}$ and r_{end} right before the leg-change, and as $\phi_{1,st} and r_{st}$ right after the leg-change, the parameters shown in Fig. 13 can be estimated by the equations below.

$$\phi_{1,st} = \phi_{1,end} - u_D \tag{21}$$

$$r_{st} = \frac{r_{end} \cos \phi_{1,end}}{\cos \phi_{1,st}} \tag{22}$$

$$\dot{\phi}_{1,st} = \frac{\dot{r}_{end}}{r_{st}} \sin u_D + \frac{r_{end}}{r_{st}} \dot{\phi}_{1,end} \cos u_D \tag{23}$$

$$\dot{r}_{st} = \dot{r}_{end} \cos u_D - r_{end} \dot{\phi}_{1,st} \sin u_D \tag{24}$$

And these equations can be rewritten, by using the state variables $(\theta_1, \theta_2, \dot{\theta}_1, \dot{\theta}_2)$ of the Mono-Bi system, as follows. Note that $\theta_m = \frac{\theta_2}{2}$.

$$\phi_{1,st} = \frac{\pi}{2} - \theta_1 - \theta_m \tag{25}$$

$$r_{st} = l_m \tag{26}$$

$$\dot{\phi}_{1,st} = -\dot{\theta}_1 - \dot{\theta}_m \tag{27}$$

$$\dot{r}_{st} = v_{lx} \tag{28}$$

By using the state variables $(\theta_1, \theta_2, \dot{\theta}_1, \dot{\theta}_2)$ obtained from the equations above, extension to walking of a supporting leg model can be realized.

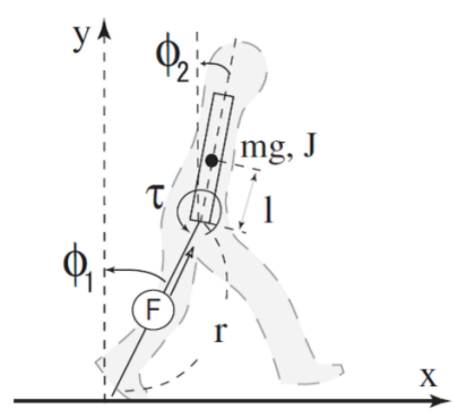


Fig. 12. The figur of the walk model

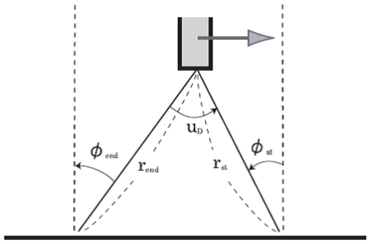


Fig. 13. The walk model at the time of change a supporting leg

C. Simulation including changing legs of walking operation

Simulation is done by placing the pole at $\omega = 50$[rad/s] for the plant in (16), and the spring coefficien of the biarticular torque τ_{12} is $K_{fix} = 550$[Nm/rad].

VI. Conclusion

We proposed the mono-bi system eliminating the monoarticular muscle pair of the second joint compared to the conventional 3 pairs 6 muscles model. Then we simplifie the kinematics of the system using a novel leg coordinates. Velocity control of COM by using this Mono-Bi system as the supporting leg model was applied to show very good tracking characteristics. Also even when the biarticular muscle τ_{12} was of a passive element, the result was fairly consistent to that of human walking. Moreover, simulations on leg-change during walking using the inverted pendulum model were done to show the feasibility of the new AFO which can reduce patients' burden.

As the next step, a relation between the reference generation and parameters of walking needs to be clarified Future work will include findin solutions to such problems, and fabrication of the prototype of the novel AFO.

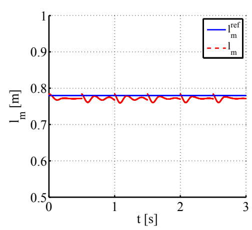
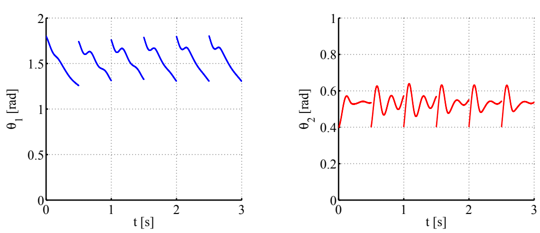
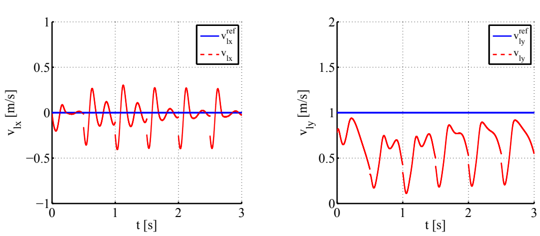


Fig. 14. Change of leg length l_m



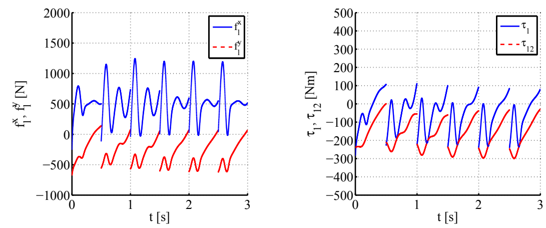
(a) First joint θ_1 (b) Second joint θ_2

Fig. 15. Joint angles θ_1, θ_{12}



(a) Velocity v_{lx} (b) Velocity v_{ly}

Fig. 16. Velocity of center of mass v_{lx}, v_{ly}



(a) Input forces f_l^x, f_l^y (b) Input torques τ_1, τ_{12}

Fig. 17. Input forces and torques in the leg coordinate

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