Modelling and control of an upper extremity exoskeleton for rehabilitation

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Abstract. This paper presents the modelling and control of a two degree of freedom upper extremity exoskeleton for rehabilitation. The Lagrangian formulation was employed to obtain the dynamic modelling of both the anthropometric based human upper limb as well as the exoskeleton that comprises of the upper arm and the forearm. A proportional-derivative (PD) architecture is employed to investigate its efficacy performing a joint task trajectory tracking in performing flexion/extension on the elbow joint as well as the forward adduction/abduction on the shoulder joint. An active force control (AFC) algorithm is also incorporated into the aforementioned controller to examine its effectiveness in compensating disturbances. It was found from the study that the AFC-PD performed well against the disturbances introduced into the system without compromising its tracking performances as compared to the conventional PD control architecture.

1. Introduction

According to the World Health Organisation’s (WHO) 2014 World Health Statistics, 8% of Malaysia’s population is well over 60 years old as of 2012 [1-2]. Furthermore, the Malaysian Ministry of Health’s annual report 2013 reported that approximately 11% and 7.2% of the Malaysian population between the ages of 0 to 18 years suffer from physical and cerebral palsy disabilities, respectively [2-3]. The report also suggests that there is an average increase of 300% of stroke patients annually. It is not uncommon that the aforesaid percentile experiences paralysis or hemiparesis (weakness or reduced strength on one part of the body) as well as cognitive disorders that affects them in performing activities of daily living (ADL)[4]. There is evidence that suggest through rehabilitation therapy, one’s mobility may be improved via continuous and repetitive training on the affected limb [2, 4-6]. Nonetheless, traditional rehabilitation therapy is often deemed both cost and labour intensive; this situation is further aggravated as the number of patients grow [4-6]. Robotics rehabilitation have been identified and engaged as a possible solution to address the shortcomings of conventional rehabilitation techniques. The use of such exoskeletons may provide longer intensive rehabilitation sessions for the patients as the presence of therapists throughout a session is no longer necessary. This would primarily reduce the fatigue level as well as enabling the therapist to tender more patients, which in turn increases more frequent treatments as well as possible cost reductions.
Hitherto, there are several robotic therapy of the upper limb that have been developed and even successfully commercialised. Upper limb robotics therapy may be demarcated into either end-effector based or exoskeleton based. The end-effector based is not attached to the patients’ whole arm but merely holds the patients’ hand or a point at their forearm in producing a particular motion. These type of robots although are relatively simpler to fabricate, it does not ensure accurate rehabilitation of a particular joint (for instance, elbow) as the resultant motion may be caused by the combination of other joints viz. shoulder and wrist joints. Amongst notable upper limb end-effector rehabilitations robots are the MIME [7], GENTLE [8] and MIT-MANUS [9].

Exoskeletons, on the other hand, are structures that resemble the human limb along its joints. Owing to its complex design, it is able to mitigate the aforementioned issue raised, therefore enabling specific muscles at specific joints to be rehabilitated. Amongst notable positional controlled upper limb exoskeletons are ARMin III [10], Rupert IV [11] and MGA exoskeleton [12]. ARMin III is a 6 degrees of freedom (DoF) exoskeleton that utilises a proportional-derivative (PD) control architecture to allow smooth positional control of a predefined angle position. Rupert IV employs a proportional-integral-derivative (PID) controller on its 5 DoF exoskeleton, whilst the 6 DoF MGA exoskeleton also made use of a PD architecture. It is worth mentioning that the aforesaid controllers demonstrate excellent performance in the absence of disturbances, nonetheless suffers considerably with its inclusion.

This study aims at investigating the joint tracking performance of a simple and robust control scheme viz. AFC-PD of a two DoF upper limb exoskeleton system subjected a number of different type of disturbances. The system intends to rehabilitate the flexion/extension of the elbow as well as the adduction/abduction of the shoulder joint in the sagittal plane. The performance of the proposed scheme will also be compared with an equivalent system that employs the conventional PD controller. To the best of the authors’ knowledge, this study is novel as the proposed controller design has yet been utilised in any upper limb exoskeleton system.

2. Dynamics and Control

2.1. Upper Limb Dynamics
The upper-extremity dynamics of both the human limb and robotic exoskeleton are modelled as rigid links joined by joints as illustrated in figure 1. The two-link model is restricted along the sagittal plane by assuming seamless human-machine interaction. This model is a rather conservative model as the frictional elements acting on both the exoskeleton and human joints as well as other unmodelled dynamics are ignored.

![Figure 1. A two-link manipulator that mimics the human upper limb.](image)

The subscripts 1, and 2 in figure 1 indicates the parameters of the first link (upper arm), and the second link (forearm), respectively. L is the length segments of the limb and the exoskeleton; \( L_c \) is the length segments of the limb as well as the exoskeleton about its centroidal axis and \( \theta \) is the angular position of the links. The Euler-Lagrange formulation is employed in deriving the equation of motions.
for the upper-extremity dynamic system. The coupled nonlinear differential equations may be written as [13]

\[ \tau = D(\theta)\ddot{\theta} + C(\theta, \dot{\theta}) + G(\theta) + \tau_d \]  

(1)

where \( \tau \) is the actuated torque vector, \( D \) is a two by two inertial matrix of the limbs and exoskeleton, \( C \) is the Coriolis and centripetal torque vector, \( G \) is the gravitational torque vector whilst \( \tau_d \) is the external disturbance torque vector. The matrix form of equation (1) is as follows

\[
\begin{bmatrix}
\tau_1 \\
\tau_2
\end{bmatrix} =
\begin{bmatrix}
D_{11} & D_{12} \\
D_{21} & D_{22}
\end{bmatrix}
\begin{bmatrix}
\dot{\theta}_1 \\
\dot{\theta}_2
\end{bmatrix} +
\begin{bmatrix}
C_{11} & C_{12} \\
C_{21} & C_{22}
\end{bmatrix}
\begin{bmatrix}
\dot{\theta}_1 \\
\dot{\theta}_2
\end{bmatrix} +
\begin{bmatrix}
G_1 \\
G_2
\end{bmatrix} +
\begin{bmatrix}
\tau_{d1} \\
\tau_{d2}
\end{bmatrix}
\]  

(2)

where the components of the inertial matrix are

\[ D_{11} = m_1L_1^2 + I_1 + m_2\left(L_1^2 + L_2^2 + 2L_1L_2\cos\theta_2\right) + I_2 \]  

(3)

\[ D_{12} = D_{21} = m_1L_1L_2\cos\theta_2 + m_2L_2^2 + I_2 \]  

(4)

\[ D_{22} = m_2L_2^2 + I_2 \]  

(5)

whereas the Coriolis components are

\[ C_{11} = -m_2L_1L_2\left(2\dot{\theta}_2\right)\sin\theta_2 \]  

(6)

\[ C_{12} = -m_2L_1L_2\dot{\theta}_2\sin\theta_2 \]  

(7)

\[ C_{21} = m_2L_1L_2\dot{\theta}_1\sin\theta_2 \]  

(8)

\[ C_{22} = 0 \]  

(9)

and for the gravitational terms

\[ G_1 = \left(m_1 + m_2\right)gL_1\cos\theta_1 + m_2gL_2\cos\left(\theta_1 + \theta_2\right) \]  

(10)

\[ G_2 = m_2gL_2\cos\left(\theta_1 + \theta_2\right) \]  

(11)

where \( m \) is the combination of both masses, whilst \( I \) is the mass moment of inertia of the exoskeleton as well as the limbs, respectively, and \( g \) is the gravitational constant taken as 9.81m/s\(^2\). The human limb parameters that are used in this study are obtained from literature [14]. The remaining parameters are listed in section 3.

2.2. Control Architecture

The Active force control (AFC) strategy is based on the principle of invariance and Newton’s second law of motion. The idea of AFC was initially conceived by Hewit and Burdess in the early eighties [15]. The effectiveness of this control strategy has been further developed by Mailah and co-researchers by integrating intelligent methods to approximate the inertial matrix of the system that is of interest. The main computation burden lies in the acquisition of an appropriate estimated inertia matrix that triggers the compensation effect against any form of disturbances [16-21].
This relatively simple yet robust control strategy has been successfully demonstrated both numerically as well as experimentally in a number of different applications [17-21]. A schematic of the AFC scheme with a PD element applied to the system is illustrated in figure 2. The PD-AFC control scheme is engaged upon the activation of the AFC loop, without its initiation, the system is controlled by the conventional PD architecture.

The torque generated is directed by the classical PD control law, typically expressed as [13]

$$\tau = K_p(\dot{\theta}_d - \theta) + K_d(\dot{\theta}_d - \dot{\theta})$$

(12)

where, $\dot{\theta}_d$ and $\dot{\theta}$ are the desired and current angular velocities, respectively, $\theta_d$ and $\theta$ are the desired and current angular positions, respectively, whilst $K_p$ and $K_d$ are the proportional and derivative constants, respectively. In order to compensate the actual disturbances, $\tau_d$, the estimated disturbance torque, $\tau_d^*$ has to be computed and is given by the following equation

$$\tau_d^* = \tau - \text{IN}\ddot{\theta}$$

(13)

where IN is the estimated inertial matrix, $\ddot{\theta}$ is the measured acceleration signal, whilst $\tau$ is the measured applied control torque.

The estimated inertial matrix adopted in the study is obtained by means of a crude approximation technique. A number of studies have shown that this method works well provided that the IN selected lies within certain bounds of the actual inertial matrix, D [17]. The inertial matrix, IN may be expressed in the following form

$$[\text{IN}] = k*[D]$$

(14)

where k is a constant bounded between $0.4 < k < 1.2$ that is proportional to the diagonal terms of D. The off-diagonal terms of the matrix are intentionally neglected. In this study, the value k of 0.4 was found to be suitable to initiate the AFC loop.

3. Results and Discussion

The simulation works of this study are performed in MATLAB and Simulink software packages. The simulation parameters employed in the simulation study are given as follows

**Upper-limb parameters:**

- Limb and exoskeleton lengths: $L_1 = 0.34$ m, $L_2 = 0.25$ m;
- Centre of mass: $L_{c1} = 0.17$ m, $L_{c2} = 0.125$ m;
- Limb masses: $m_{limb1} = 1.91$ kg, $m_{limb2} = 1.22$ kg;
- Exoskeleton masses: $m_{exo1} = 0.34$ kg, $m_{exo2} = 0.25$ kg;

Mass moment of inertia of limb: $I_{limb1} = 0.2374$ kg.m$^2$, $I_{limb2} = 0.0873$ kg.m$^2$;

Mass moment of inertia of exoskeleton: $I_{exo1} = 0.0131$ kg.m$^2$, $I_{exo2} = 0.0052$ kg.m$^2$.
Controller parameters:
Controller gains (obtained heuristically):
$K_{p1} = 3000$, $K_{d1} = 100$;
$K_{p2} = 800$, $K_{d2} = 50$;

Diagonal elements of estimated inertia matrix:
$IN_1 = 0.2935 \text{ kg.m}^2$, $IN_2 = 0.0743 \text{ kg.m}^2$.

Simulation parameters:
Integration algorithm: ode2 (Heun)
Simulation start time: 0.0
Simulation stop time: 10 sec
Fixed-step size: 0.001

A sinusoidal input with an amplitude of 45° (0.7855 rad) is fed into the system at both joints to investigate the tracking performance of both the conventional PD and the PD-AFC control strategy. This input replicates the flexion/extension and adduction/abduction exercise at the elbow joint (joint 1) and shoulder joint (joint 2), respectively. In order to investigate the robustness of both control architectures, two types of disturbance are included in the system, viz. a constant 100 N.m. disturbance torque and a 500 N.m. harmonic disturbance torque. Figures 3 to 5 depicts the results obtained from the simulation study.

**Figure 3(a).** Result of joint 1 angle trajectory without any disturbance.

**Figure 3(b).** Result of joint 2 angle trajectory without any disturbance.
Figure 3(c). Tracking error of joint 1 without any disturbance.

Figure 3(d). Tracking error of joint 2 without any disturbance.

Figure 4(a). Result of joint 1 angle trajectory with a constant disturbance of 100 N.m.
The joint root mean square error (error_{RMS}) for the system that is regulated by the PD-AFC controller without the inclusion of any form of disturbance is 2.131 mrad at joint 1 and 1.913 mrad at joint 2. The results suggest that the PD-AFC control scheme is far more superior than the conventional PD architecture as the latter manage to reduce the error_{RMS} to only 6.694 mrad and 9.198 mrad at joint 1 and joint 2, respectively. The effectiveness of the PD-AFC architecture in rejecting disturbances is demonstrated in figures 4 to 5. It is evident from the figures that the proposed control scheme (PD-AFC) is able to compensate any form of disturbances (harmonic and constant) effectively whilst...
maintaining excellent joint tracking error as compared to its traditional counterpart. Table 1 below summarises the errorRMS for all cases.

Table 1. Summary of joint root mean square tracking error (errorRMS).

| Disturbance Type   | Elbow joint, $\theta_1$ errorRMS (mrad) | Shoulder joint, $\theta_2$ errorRMS (mrad) |
|--------------------|----------------------------------------|--------------------------------------------|
|                    | PD                                     | PD-AFC                                     |
| None               | 6.694                                  | 2.131                                      |
| Constant (100 N.m.)| 32.675                                 | 2.130                                      |
| Harmonic (500 N.m.)| 114.958                                | 2.132                                      |

4. Conclusions

It is apparent from the study that the proposed PD-AFC performs reasonably well even under the influence of external disturbances. The conventional PD control strategy provides satisfactory tracking performance without the presence of disturbance, however, suffers considerably upon the onset disturbances. The study further implies the effectiveness of the proposed controller for the early stage of upper limb rehabilitation. Further investigation may be carried out by subjecting the system to other form of disturbances as well as incorporating intelligent methods (fuzzy logic, neural network, etc.) in acquiring the suitable estimated inertial matrix.

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